Even Numbered Posters
Wednesday, March 11th 4:00 - 5:00pm

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Effect of Muscle Reflex of Lower Limbs on Injury of Knee

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INTRODUCTION

Accidental falls due to a minor impact on the lateral knee would be dangerous for humans. The minor impact could be defined that measured impact acceleration at the medial knee is less than 100 m/sec^2. It is known that the impact acceleration could not cause a serious injury on the lower limbs [1]. However a report indicates that a high frequency, or rate loading, such as an impact, develops acute inflammatory responses in the ligaments although no rupture was in evidence [2]. Since an acute inflammation could result in a neuromuscular disorder, the knee injuries due to the minor impact would have significant importance on the chronic knee pains associated with minor ligament injuries of the medial collateral (MCL) and anterior cruciate (ACL) ligaments. In the previous study [3], a distinct reflexive contraction of the hamstrings of felines and humans was observed to reduce the tensile force of the knee ligaments when the tension of ACL and MCL was increased. It is a hypothesis of this study that the ligamento-muscular reflex in the lower limbs affects the injury of the knee during a minor impact on the lateral knee of human. The purpose of this study is to analyze effects of involuntary muscle activations of lower limbs on injury of the knee when the lateral knee of human in the mid-stance is impacted at the velocity of .2.88 km/hour.

CLINICAL SIGNIFICANCE

The reflexive knee motions of humans by involuntary muscle activations were observed during the minor impact on the lateral knee. Based on dynamic simulations, the effects, of involuntary muscle activations in the lower extremities of human due to the minor impact on the lateral knee, could be beneficial to reduce the risk of injury of ACL and MCL.

METHODS

Electromyography (EMG) of the prime movers in the low extremities was measured from maximum voluntary muscle contraction (MVC) tests using five young subjects. Five muscles from the thigh (rectus femoris, vastus lateralis, vastus medialis, semitendinosus, and biceps femoris), and three muscles from the shank (tibialis anterior, and medial and lateral gastrocnemius), were selected, and the EMG was measured (Noraxon, USA). Isometric muscle exercises were performed at 33˚ and 17˚ of knee angle, for the both the right and left legs respectively using a servo-electrical exercise machine (Biodex Medical System, USA). 33˚ and 17˚ are the flexion angles of the knee joint of the advanced, and following, leg when the gait of a human is in the opposite toe off condition. The measured MVC EMG signals were used in normalizing the EMG data from the falling tests. For the falling tests, a specially fabricated microprocessor controlled sled was utilized to simulate the minor impact at the velocity of .2.88 km/hour. Activation time sequences and temporal EMG data were measured during the falling tests. In addition, kinematics of low extremities and pelvis were measured using a 3D motion capture system (Motion Analysis, USA). The same subjects used in the previous experiments participated in the tests. In addition, the EMG electrodes were instrumented as in the previous experiments. Impact accelerations at the medial knee were measured to validate the developed finite element (FE) human model during the falling tests.

To ensure the safety of the subjects, a rehabilitation doctor was in attendance at the tests. Figure 1 shows the experiment. Then, simulations for the minor impact have been performed using a FE human model with muscles in the lower extremities illustrated in Figure 2. The long bones and soft tissues in the lower extremities of the dummy were modeled to be deformable FE with Hill type muscles. The other parts were modeled as ellipsoidal rigid bodies. The hip, knee, and ankle joints were modeled as a spherical, revolute with a translation, and universal joint, respectively. The previously obtained sequence of the muscle activations, and the normalized temporal MES by values of the MVC MES, was used as the input parameters for the FE human model during the minor impact simulations. The impactor representing a car bumper was modeled using FE. Then, the same simulation setups and parameters as used in the previous impact tests were applied to the simulations. The simulations (MADYMO V6.2.1, TNO, Netherlands) were performed for two cases, i.e., with and without muscle activations, to understand effects of involuntary muscle reflex in the lower limbs on injury of the knee during the minor impact.

RESULTS
Figure 3 shows the predicted temporal impact accelerations at the medial knee, with and without muscle activation in the low extremities, when the impact velocity was 2.88 km/hour. The simulated maximum accelerations and knee extension moments with and without muscle activations were 17.2 and 21.5 m/sec^2, and 44 and 50 Nm, respectively.

DISCUSSION
The predicted temporal impact acceleration with muscle activations was found to be relevant within the experimental results shown in Figure 4. Also, a similar motion sequence of the human model to that of subjects was predicted during the falling event. In this study, fast knee flexion motion and co-contraction of the quadriceps, in low level activity by the reflex just after the onset of impact were observed. This could be a protective mechanism for the ACL and MCL injuries as indicated by the previous studies [2, 3]. Reductions of the maximum values in the impact acceleration and knee extension moment for with muscle activations could support the mechanism.

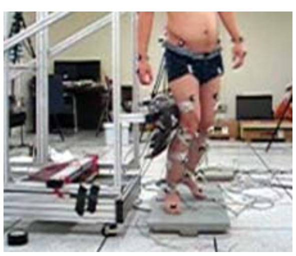

Fig. 1 Experiment

Fig. 2 FE Model

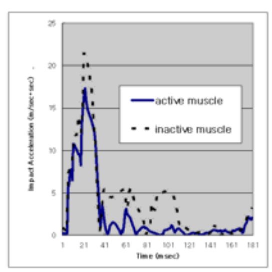

Fig. 3 Predicted Accel.

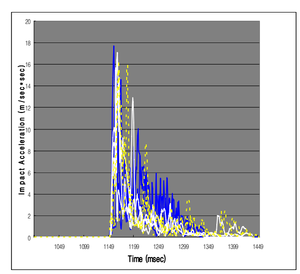

Fig. 4 Measured Accel.

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Kinematic Gait Analysis after Total Knee Arthroplasty Comparing a Mobile-Bearing Implant to the Normal Contralateral Knee

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Introduction

Total knee arthroplasty [TKA] has become a standard of care in treating end stage degenerative and inflammatory arthritis. The fixed-bearing knee implant is commonly used to provide pain relief while maintaining the patient's preoperative lifestyle and function, and many gait studies have shown these benefits [1, 2, 3]. However, patients with fixed-bearing knees may possibly retain gait abnormalities due to its non-conforming design and limit a patient's function. Mobile-bearing total knee arthroplasty is designed as a dual articulation to enable conformity in its kinematics and allow for closer recapitulation of normal knee movement. Our hypothesis was that the mobile bearing, rotating platform TKA accommodated to each individuals normal gait kinematics. This study seeks to understand the kinematic differences in gait between the rotating platform TKA and the contralateral normal knee.

Clinical Significance

The functionality of the knee is important to retain after total knee arthroplasty. Mobile-bearing knees may help reproduce normal knee movements and may eliminate the gait deviations seen in the fixed-bearing knee population. The conforming mobile-bearing design may provide higher patient satisfaction.

Methods

Six males at least one year status post unilateral mobile-bearing TKA underwent computerized motion analysis (VICON Nexus Motion Analysis System, Oxford Metrics, Oxford, England) while performing normal gait movements. The mobile-bearing knee models were all Depuy PFC Sigma Rotating Platform TKA's All subjects had no other joint involvement or symptoms based on patient medical history and review of systems assessment. The mobile-bearing knees were tested against the natural contralateral knees of the six subjects, as well as compared with a group of previously collected normal adult knees. Standard statistical methods were used in calculating results.

Results

No statistically significant differences were found among the groups in flexion/extension during any part of the gait cycle. The maximum flexion achieved in swing phase for the mobile-bearing knee was 58.0° ± 1.4 compared to 53.9° ± 6.0 and 57.0° ± 19.4 for the contralateral and normal knee groups respectively. Differences were found in knee varus/valgus for both stance and swing phases although the difference were not statistically significant. The mobile-bearing knee showed an average of 1.1° ± 3.8 of valgus during stance and 1.3° ± 3.5 of valgus during swing. The contralateral knee showed 2.0° ± 2.5 of varus in stance and 1.2° ± 2.4 of varus in swing and the normal knee group showed 1.1° ± 2.2 of varus in stance and 2.8° ± 2.2 of varus in swing. Figures 1 shows the knee varus/valgus for the three groups during stance and swing. The knee rotation showed external rotation of all three groups with the mobile-bearing average of 6.1° ± 14.1 during stance and 9.8° ± 13.6 during swing. This average is not significantly different from the contralateral group or the normal group with averages of 1.5° ± 12.2 and 10.6° ± 1.5 in stance respectively and 3.6° ± 13.6 12.3° ± 3.4 in swing respectively. Figure 2 shows the external rotation for the three groups during stance and swing phases.

Figure 1

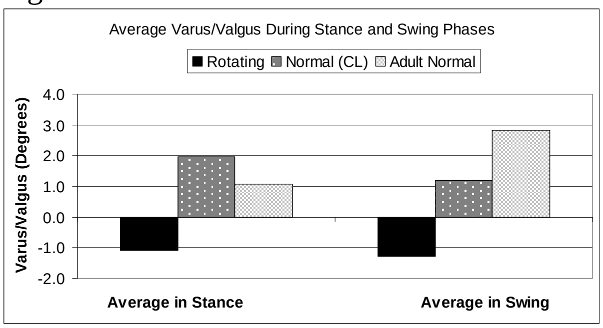
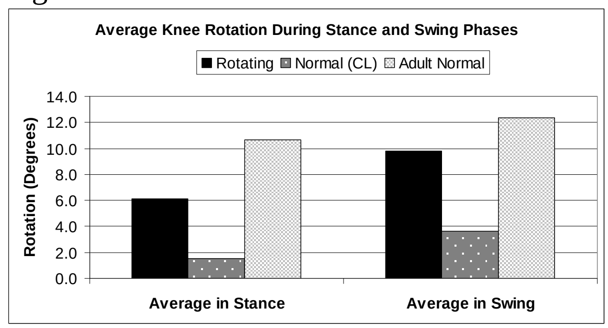


Figure 2



Discussion

In flexion, the mobile-bearing knee implant achieves the same range of motion as both the contralateral knees and the normal adult knee indicating full function in the sagittal plane is retained post surgery. In varus/valgus, the mobile-bearing knee implants tend to have a slight varus thrust whereas both groups of normal knees have slight valgus thrusts. This is not a statistically significant difference and is may be due to the design of the implant or surgical technique. The knee rotation was only slightly decreased from normal adult knees but was increased over the contralateral knee group. This may indicate that the mobile-bearing platform allows for more rotation replicating the normal knee. While some differences and trends have been shown here the results are limited by the low number of subjects and the variability between subjects. In order to obtain more significant results more subjects will be tested. Additionally, a set of fixed-bearing knee subjects will be tested to more thoroughly characterize mobile-bearing and fixed-bearing implants with regards to gait analysis.

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Acknowledgements

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RECTUS TRANSFER PATIENTS SHOW IMPROVED DYNAMIC BALANCE POST-SURGICALLY

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INTRODUCTION

A number of parameters have been utilized to quantify success of rectus-transfer outcomes including improved knee flexion [1], decreased energy expenditure [1][2], and decreased frequency of tripping [2] post-surgically. CP patients generally have impaired balance compared to their matched normal peers [3], yet evaluation of specific improvement in balance due to rectus transfer surgery is not routine.

CLINICAL SIGNIFICANCE

This research illustrates that rectus transfer patients also improve their dynamic balance scores as a result of the surgery and that the resulting scores are similar to their normal age-matched peers . This evidence of improved dynamic balance of patients as a result of the rectus-transfer surgery illustrates another benefit of the surgery which can be quantified using gait analysis.

METHODS

Retrospective data from eighteen patients who had undergone multi-level surgeries including rectus transfer were randomly selected (ages 10.0 ±4.6 years). Balance indices (K_N and D_N) were calculated according to Niiler et al [4] for both pre and post surgical trials. Descriptive statistics including averages and standard deviations were calculated on K_N and D_N for each trial. Pre to post statistics were related using regression analysis and Welch's two sample t-test. In the case of the t-test, it was necessary to transform data using the square-root function in order to obtain normalcy before conducting the t-test.

RESULTS

T-test results indicated significant differences ($p < 0.002$) only in the average D_N with D_N decreasing by roughly 50% from pre to post surgery. Regression of post-surgical averages of D_N versus pre-surgical averages were significant ($p < 1e\text{-}6$) and resulted in the following relationship between pre and post surgical averages.

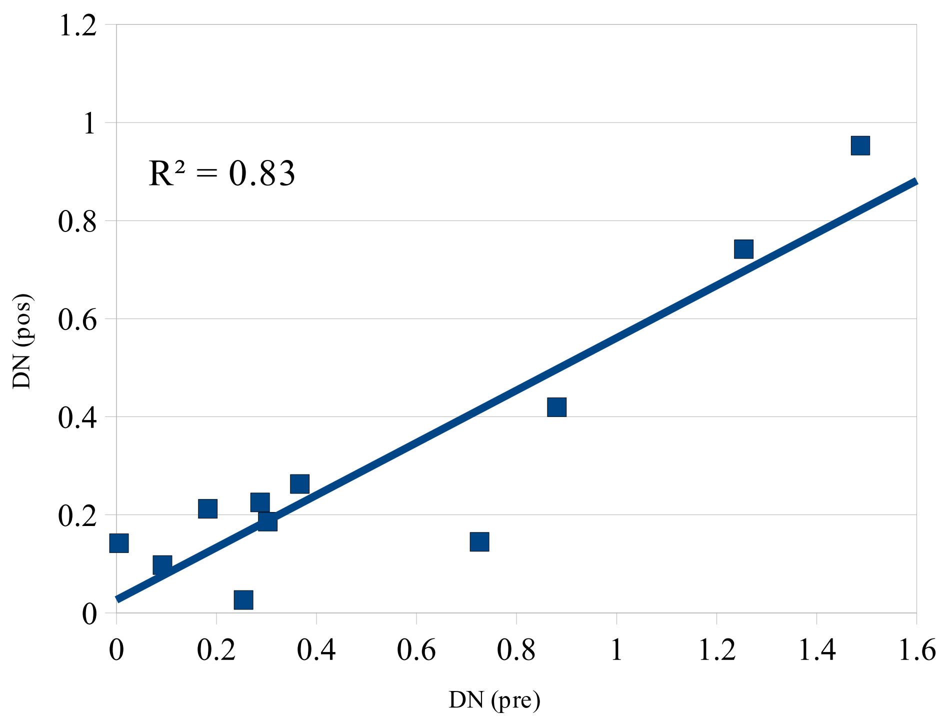
$$D_N(\text{post}) = 0.52887\, D_N(\text{pre}) - 0.01229$$

Table I: Pre- versus post-surgical averages for balance indices. Significant differences between pre and post conditions are indicated by an asterisk.

	Avg(D_N)*	Std(D_N)	Avg(K_N)	Std(K_N)
Pre-Surgery	0.313	0.539	0.230	0.383
Post-Surgery	0.153	0.433	0.253	0.409

DISCUSSION

The balance index D_N is a normalized virtual moment arm between the center of mass (COM) of the body and the line of support (LOS) joining the feet at any time [4]. As D_N exceeds unity, the COM deviates from the LOS more than a foot width, and the subject is likely to become more unstable. The results of this study show that the average D_N across a gait cycle decreases almost by a factor of two as a result of rectus transfer surgery indicating improved balance. Further analysis of pre- to post- surgical velocities indicate no significant differences implying that the decrease in D_N is not due to more conservative movement due to recovery. Comparison with lab normals across a similar age group ($\langle D_N \rangle$(normals) $= 0.106 \pm 0.110$) indicated that post-surgically, the CP patients were approaching their normal peers in positioning of their COM. The K_N balance index is more related to changes in the position of the COM and is thought to be indicative of motor control [4]. Rapid corrections to the position of the COM result in a higher K_N. The lack of change of K_N as a result of surgery is expected since the rectus-transfer does not treat any of the underlying causes of the spasticity.



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Intramuscular Rectus Femoris Lengthening for the Correction of Stiff Knee Gait in Children with Cerebral Palsy: An Evaluation using 3D Motion Analysis

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Introduction: Gait analysis has been important in understanding the possible causes of "stiff knee" gait. This gait pattern is characterized by reduced peak knee flexion in swing phase which results in interference with foot clearance and reduced overall sagittal plane knee motion[6]. The most widely accepted treatment to address stiff knee gait is rectus femoris transfer[5]. The proposed mechanism of this procedure is the reduction of the knee extensor effect of the rectus femoris as well as the conversion of the muscle into a knee flexor in swing. Recent studies, however, have posed challenges to the proposed mechanism of rectus femoris transfer[2,7]. The benefit of rectus femoris transfer may be primarily attributable to the muscle's reduction as a knee extensor rather than its conversion to a knee flexor in swing. The purpose of this study was to evaluate an alternative procedure for the treatment of stiff knee gait, the intramuscular rectus femoris lengthening (RFIL) which would reduce its function as a knee extensor.

Statement of Clinical Significance: Rectus femoris intramuscular lengthening theoretically reduces the knee extensor function of the rectus femoris. It involves less surgical dissection and less operative time than rectus femoris transfer. If the RFIL is shown to be as or more effective in addressing sagittal plane knee motion as the rectus femoris transfer, it could provide a less invasive and less morbid surgical option in the treatment of stiff knee gait.

Methods: This study was a retrospective data review of ambulatory patients with a diagnosis of cerebral palsy who had RFIL. The indications for rectus femoris intramuscular lengthening were identical to those of rectus femoris transfer and include kinematic, electromyographic and clinical findings. Patients must have had preoperative and postoperative comprehensive gait analyses. Three-dimensional kinematic and kinetic data were collected as part of the standard of care using a VICON 512 motion measurement system (VICON Motion Systems, Inc Lake Forest, CA) following standard techniques[4]. A representative trial was selected for analysis both pre and postoperatively. Pre to postoperative differences were measured using a Student *t* test ($p < 0.05$). Selected sagittal plane kinematic and kinetic parameters were analyzed.

Results: A total of 43 patients (72 sides) treated between 1991 and 2008 with preoperative and postoperative gait analyses following RFIL were analyzed. Additional procedures included femoral varus derotational osteotomy (14 sides), femoral derotational osteotomy (8 sides), hamstring lengthening (61 sides), psoas lengthening (5 sides), adductor lengthening (9 sides), gastroc-soleus/tendoachilles lengthening (32 sides), tibial derotational osteotomy (17 sides), and posterior tibial tendon lengthening (2 sides). The mean age at surgery was 9.0 years (SD +/- 4.3) and the mean time after surgery at postoperative gait analysis was 18.7 months (range 7-54 months). There were

27 males and 16 females. The results of selected sagittal plane knee parameters analyzed for this cohort are summarized in the accompanying table.

Table—Comparison (mean +/- 1 SD) sagittal plane knee parameters pre vs. post surgery, N=43 (72 sides)

Parameter	**Reference**	**Preop**	**Postop**	***P* (pre-post)**
Peak knee flex (SW) deg	66.1 (7.4)	52.9 (11.4)	51.9 (9.0)	0.497
% GC peak knee flex (SW)	71.4 (2.1)	81.9 (5.5)	79.5 (4.9)	**0.001**
Knee sagittal ROM (entire GC) deg	63.1 (7.1)	36.6 (12.3)	39.3 (13.6)	**0.042**

(Legend: SW = swing, ROM = range of motion, GC = gait cycle, ST = stance)

Discussion : The most common sagittal plane knee parameters evaluated when assessing treatments for stiff-knee gait are peak knee flexion in swing, timing of peak knee flexion in swing, and knee sagittal range of motion[1,3]. When comparing pre- and postoperative gait analysis data, our cohort showed maintenance of peak knee flexion in swing, improved timing of peak knee flexion in swing (occurs earlier in the gait cycle), and increased knee sagittal range of motion (related to hamstring procedure). This is comparable to previously published results evaluating rectus femoris transfer for the treatment of stiff knee gait[1,6].

The data suggest that the RFIL diminishes the knee extensor function of the rectus femoris and thus, offers an alternative to rectus femoris transfer in the treatment of stiff knee gait. The RFIL shows similar benefits to rectus femoris transfer in terms of maintaining peak knee flexion in swing, improving timing of peak knee flexion, and increasing knee sagittal range of motion at 18 months post surgery[1]. The procedure itself is technically less demanding with less surgical dissection involved. With less surgical dissection, patients have less postoperative pain, are able to be mobilized more easily, may participate in rehabilitation earlier and more vigorously, and thus the overall morbidity of the procedure is less than that of rectus femoris transfer. In the future, we intend on bringing these patients back to assess longer-term outcomes.

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Gait Analysis for Conservative/Surgical Treatment Planning in an Adult with Traumatic Brain Injury and Orthopedic Complications : A Case Study

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Patient history: M. M. was a 37 year old male with a 16-year history of traumatic brain injury who was referred for recommendations to improve his walking function. He also had a left ankle fracture (12 years prior), a right total knee joint replacement (14 months prior), and complaints of low back pain. M. M. ambulated without assistive devices or orthoses except for shoe-insert orthotics for his low back pain. He reported performing daily lower extremity stretching and 3 sessions per week of strengthening exercises and treadmill walking. Gait deviations and clinical findings were similar bilaterally, but more impaired on the left than the right. This case study is focused on the left lower extremity.

Clinical Data: Relevant clinical findings included limited passive range of motion in ankle dorsiflexion (-10 degrees with rear foot inversion and knee extension, +10 degrees with knee flexion and eversion), and straight leg raising (55 degrees) indicating hamstring tightness. Muscle strength was reduced in the ankle plantar flexors (Fair) and hip extensors (Good minus). Strength was Normal by manual muscle testing in hip flexor, knee extensor, ankle dorsiflexor, invertor, and evertor muscle groups. Moderate spasticity was present in the plantar flexors and hamstrings.

Gait Data: He walked at 50 meters/minute (56 % of normal) with reduced stride length (65% of normal) and decreased cadence (86% of normal). Quantitated motion analysis of the left lower extremity identified primary gait deviations in stance including excessive ankle dorsiflexion (20 degrees), knee flexion (25 degrees), and thigh flexion (peak extension at 5 degrees) (Figure 1). Marked rear foot eversion (20 degrees), knee valgus (10 degrees), and knee external rotation (25 degrees) were recorded throughout. In swing peak knee flexion was decreased (45 degrees) with two abnormally timed peaks at toe-off and in late mid swing. Knee extension was decreased in terminal swing with initial contact in 20 degrees of flexion. Kinetic analysis identified excessive subtalar eversion, and knee flexion and valgus moments in single limb stance with near normal ankle dorsiflexion moment despite his plantar flexion weakness indicating reliance on passive support from excessive dorsiflexion to the limits of passive mobility. Ground reaction forces in loading were excessive during walking in shoes.

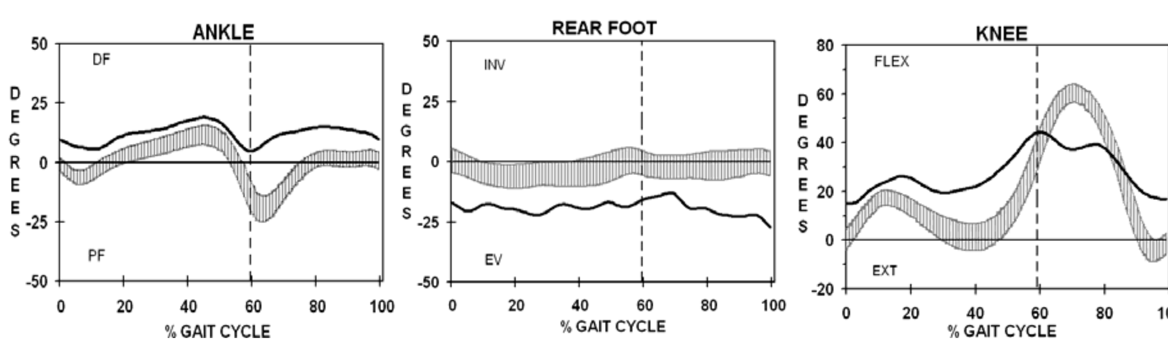

Figure 1. Kinematics of the left lower extremity during free walking

Dynamic EMG recording during free barefoot walking identified premature and vigorous activity in soleus (SOL) from initial contact through stance. Gastrocnemius (GAS) had a similar pattern, but low intensity. The strong SOL activity was attempting to control dorsiflexion and eversion, while GAS was reduced because of excessive knee flexion. Posterior tibialis and peroneus brevis had a brief burst of activity in loading with quiescence until terminal stance indicating poor control of the rear foot eversion in mid stance. Peroneus longus was not active during walking despite moderately vigorous activity in muscle testing both before and after the walking trials. Thus, the primary contributor to rear foot eversion in single limb stance was passive progression of body weight to the limits of both dorsiflexion and eversion which permitted dorsiflexion of the mid foot. Gluteus Maximus (GMAX) activity was insufficient with very low intensity and biceps femoris long head was low and out of phase. Consequently, the primary hip extensor muscle was semimembranosus (SMEM) which had vigorous, nearly continuous activity throughout stance. While SMEM provided support at the hip and pelvis, its activity also exacerbated his knee flexion in stance.

Treatment Decisions and Indications: Decreased passive dorsiflexion mobility in the ankle joint resulted in subtalar eversion with progression of body weight over his foot. Excessive subtalar eversion in stance unlocked the mid tarsal joint and allowed increased dorsiflexion mobility in the mid foot. Calf weakness removed the tibial restraint during stance resulting in excessive dorsiflexion and knee flexion. Recommendations to improve foot posture and knee motion: 1). Bilateral poly articulated ankle foot orthoses (AFO's) with free plantar flexion and a dorsiflexion stop; and 2). University of California Biomechanics Laboratory orthoses (UCBLs) with medial arch support worn inside the articulated AFO. The UCBL would control eversion and the dorsiflexion stop would support his weak calf and provide support to the tibia allowing the knee to extend. Since his calves were weak bilaterally with excessive knee flexion and eversion in stance, he would benefit from bilateral bracing.

If the bracing did not correct his excessive knee flexion in stance, temporary injection to the SMEM (lidocaine or botox) was recommended to determine if reduction of the SMEM decreased knee flexion without making the hip unstable. If this was successful, either repeated botox injections or surgical lengthening of the SMEM could be considered. If neither bracing nor injection of SMEM corrected his knee flexion, the lateral third of the Achilles tendon could be folded medially to reinforce the medial control of the ankle and subtalar joints. Additional recommendations included: 1) continue hamstring stretching, 2) discontinue calf stretching exercises because he had stretched his foot into excessive eversion and mid foot dorsiflexion.; 3) increase the intensity of hip extension exercises and perform them in knee flexion to target the GMAX; 4) Wear soft-soled shoes to reduce the impact forces of loading. His low back pain is likely exacerbated by his high ground reaction forces.

Summary: Typically, excessive dorsiflexion and knee flexion is controlled with a dorsiflexion stop AFO. The excessive eversion, however, makes bracing more complicated. An articulated brace is preferable because a solid orthosis causes a knee flexion thrust in loading when the ankle is not allowed to plantar flex. The foot plate of an articulated AFO cannot be made narrow enough to adequately contour to the calcaneus and support the subtalar joint and still accommodate the metal joints of the brace. The contoured UCBL inside the AFO is able to fit closely around the calcaneus.

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Use of lower extremity electromyography to evaluate changes in postural control four years after acquired brain injury

Introduction: There is limited evidence to support and guide rehabilitation of motor control following acquired brain injury (ABI). Outcomes are thought to be dependent on numerous factors, although progress is often difficult to evaluate owing to the complex nature of the condition. Traditionally, rehabilitation efforts have been focused on the initial period following ABI. However, as time passes, provision of therapy services typically diminishes, commensurate with expectations for recovery and assumptions about rehabilitation potential.

Patient history: The participant in this case study was a 14-year-old girl who sustained an ABI, resulting in diffuse axonal injury, four years earlier. Clinically she demonstrated cognitive and speech impairments, as well as severe motor difficulties. She used a power wheelchair for mobility, and was able to perform standing transfers with assistance. The participant's goal in physical therapy sessions was to begin to use a walker.

Clinical data: The participant displayed moderately severe hypertonicity in her left leg, and wore an ankle foot orthosis on her left side. She was able to stand with support, bearing minimal weight on her left leg, in a position of excessive left hip and knee flexion and ankle plantarflexion. Voluntary motor control of her left leg was severely impaired, and she lacked the ability to voluntarily extend her hip and knee. During the intervention period, weekly physical therapy appointments focused on assisted standing activities and neuromuscular electrical stimulation to her left rectus femoris (RF) and vastus medialis muscles. After twelve weeks, she was able to begin using a walker during therapy sessions.

Electromyographic data: Surface electromyography (EMG) was used at the beginning of the intervention period, and after 2 months and 7 months of weekly therapy, in order to evaluate the activation of RF and biceps femoris (BF) muscles during assisted standing. On each occasion, EMG activity was recorded during ten trials of 60 seconds each, as the participant stood with the assistance of one person (as required to maintain standing). Electrodes were positioned in a bipolar configuration over the BF and RF muscles according to SENIAM (Surface Electromyography for the Non-Invasive Assessment of Muscles) guidelines. The EMG signal was sampled at 2000 Hz and band pass filtered between 30 and 500 Hz. Recordings were evaluated visually, and mean amplitudes (normalized to peak EMG) for each muscle were compared over time.

The data showed an increase in mean EMG activation of the left BF and RF over time (Figure 1). Visual inspection of the EMG recordings suggested that recruitment of all four muscles was more symmetrical at the 2 month reassessment, an apparent change from pre-test recordings in which a reliance on the right RF muscle for postural adjustments was demonstrated. In addition, the participant was able to stand independently for several seconds during the 2 month reassessment, which had not been possible during the pre-test.

From the 2 month to the 7 month reassessment, mean EMG activation levels remained similar for all muscles except right BF. However, between these two reassessments, qualitative motor control gains continued to be evidenced in her ability to voluntarily extend her knee, and in her use of the walker. For example, she was able to walk 50 feet 3 months after the initial assessment, and 750 feet at the 7 month reassessment.

Treatment Decisions and Indications: Taken together, the EMG data and clinical observations help to quantify this adolescent's changes in lower extremity postural control during standing, four years after ABI. The results were interpreted to support continued regular therapy intervention, although current conventional practice guidelines may have recommended otherwise, given the length of time since her injury. They also provide some insight into the neuromuscular and postural control strategies used by the participant over the course of the intervention period. It should be noted that EMG recordings must be interpreted carefully in the presence of spasticity, and when testing occurs on different days, due to issues of reliability. However, in light of the accompanying clinical observations, the data suggest that the observed changes in muscle activation provide valid adjunctive information to inform and evaluate clinical outcomes.

Summary: The results of this study demonstrate that clinically significant gains in motor control and functional performance can continue for several years following ABI. The etiologies of muscle weakness and neuromuscular control difficulties in ABI are poorly understood, particularly as they relate to functional outcomes such as gait and postural control. However, EMG plays a unique role in the examination of neuromuscular recruitment, and this case study illustrates its novel application in a clinical environment. With appropriate interpretation, EMG offers the clinician a unique, objective means of quantifying changes in neuromuscular recruitment that may not be evident from routine clinical outcome measures alone. This information can provide an additional means to justify and inform decisions about therapy service provision. Future research is needed to explore a means of identifying clinically significant changes in EMG activation, in order to provide more reliable information to assist clinicians in problem-solving and monitoring patient progress.

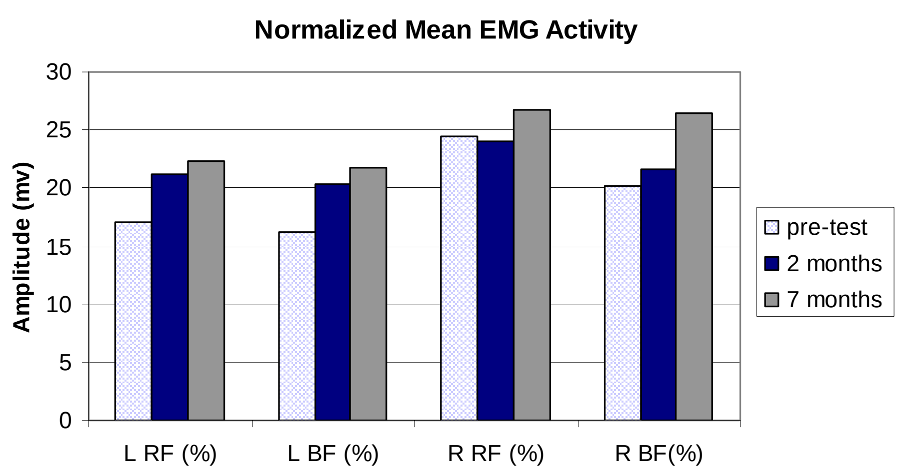

Figure 1. Graph showing normalized mean EMG activity for right (R) and left (L) rectus femoris (RF) and biceps femoris (BF) muscles during assisted standing. Measurements were recorded at the start of the intervention period (pre-test), and 2 and 7 months later. Activity of the left BF and RF muscles increased during the intervention period.

OPEN CHAIN FOOT CONTROL IS A FUNCTION OF FOOT STRUCTURE DURING RUNNING

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INTRODUCTION

Running involves rapid loading of the body's structure as it moves from a projectile in space to an actively controlled system in stance. Several works demonstrate that feedback loops exist, but are too slow to control the higher forces experienced at impact loading (1). This compromise is reportedly due to the short time in which contact occurs (2) and also due to down-regulation of the motor-neuron pool at landing (3). Therefore Loeb coined the term "preflexes" to describe the alterations in a neural control system that occur to influence joint stability (4). This component allows the individual to anticipate or tune joint stabilization forces based on anatomical structure (5). Understanding of the proprioceptive nature of the foot allows us to better understand its function. Function is believed to be somewhat dependent on structure, however the proactive orientation of the foot has not been identified during running. The goal of the authors is to describe the open chain orientation of the foot prior to contact and examine variability between foot types.

CLINICAL SIGNIFICANCE

Open chain orientation has been of interest recently to observe various aspects of gait and pathology (6), but simple ankle/foot models have not allowed examination of foot structure during running. Examination of foot structure warrants attention to determine if there is a relationship between mobility in the foot and its preparation for contact in running. Analysis and knowledge of these strategies may allow increased specificity into a patient's rehabilitation program, athletic training programs, injury pathomechanics, and footwear design.

METHODS

72 runners were screened by a physical therapist to clinically assess navicular drop for establishing foot type. The differential between navicular height in the relaxed calcaneal position while sitting and standing was recorded as functional navicular drop(7). Classifications based on this functional drop were established to group the subjects as hypomobile (0-3 mm), neutral (4-6 mm), and hypermobile (7+ mm). Kinematic data were obtained during barefoot running using a 10 camera Vicon 624 motion analysis system (Vicon Peak, Lake Forest, CA, USA) and an instrumented treadmill. All data was captured at the runner's self-selected shod pace. Retro-reflective markers were placed on the following anatomical landmarks of each foot: lateral malleoli, medial and lateral calcanei, medial first and lateral fifth metatarsal heads. Metatarsal and calcaneal angles were calculated as the angle between the horizontal plane of the laboratory coordinate system and the line connecting the medially and laterally placed markers of the forefoot and rearfoot, respectively. It is important to note that the height of the medial and lateral metatarsal and calcaneal markers were placed identically so that metatarsal and calcaneal angles were zero for each subject during normal standing. Average curves for these angles were created from 10 consecutive cycles of gait for each subject. Open chain

metatarsal and calcaneal inversion were calculated as the magnitude of the respective angle at 99% of the gait cycle. Between foot type comparisons were made using independent-samples t-tests. Within each foot type, paired-samples t-tests were used to compare metatarsal inversion to calcaneal inversion. Statistical significance was defined at $p<0.05$.

RESULTS

This study observed differences in foot orientation in the frontal plane just prior to contact (Table 1). The rigid structure of hypomobile runners exhibited a near identical inversion angle in both the metatarsals and calcaneus. Neutral runners were observed to land with slightly less inversion at the calcaneus and slightly more inversion at the metatarsals when compared to the hypomobile group, although both differences were not statistically significant. While neutral runners did not exhibit significance in mean difference, they trended toward greater inversion at the metatarsals than at the calcaneous. When compared to the neutral group, hypermobile runners landed in a significantly increased calcaneal inversion, and trended to an increase over the hypomobile group. The hypermobile group revealed significantly increased metatarsal inversion from the values observed in both the hypomobile and neutral runners. Additionally, metatarsal inversion and calcaneal inversion were significantly different within the hypermobile foot type.

Table 1. Inversion projections at contact

	n	Calcaneous	Metatarsal	Mean Difference	p
Hypo	24	7.6 ± 2.8	7.6 ± 3.9	0.0 ± 4.4	0.99
Neutral	26	6.9 ± 3.0	8.4 ± 3.2	1.4 ± 3.7	0.06
Hyper	22	*9.2 ± 2.8* †*	*11.7 ± 4.7* †‡	*2.5 ± 5.0*	*0.03*

† Statistically significant difference between Neutral and Hyper (p<0.01)

‡ Statistically significant difference between Hypo and Hyper (p<0.01)

* Statistical trend between Hypo and Hyper (p=.05)

CONCLUSION

The study reveals that different foot types do demonstrate subtle alterations in pre-contact orientation that are likely respective to each foot type to prepare the body for stance. While structural orientation may guide the path of the body's range of motion, our results indicate that an individual's contact style is likely modified to alter the demand on passive and active structures in anticipation of foot contact in running.

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CHANGES OF DIMENSIONAL PROPORTIONALITY AND ITS EFFECT ON FEET DIMENSIONS OF TEENAGER GIRLS

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Introduction

Regular foot measurements for industrial shoe production are the main condition to provide shoes with a good fit for the feet. A detailed knowledge of shape changes in children feet during growing is an important issue to design footwear with comforts and healthy attributes for population [1].

In literature and shoe business, it is often assumed that the ball girth is 2.5 times the ball width. This assumption is based on 2-Dimensional foot data and on empirical research. In the study of Bax and Smiths a correlation was found between the ball girth and the ball width for an adult male population using 3- Dimensional foot data [2].

The studies performed in Tubingen stress the fact that, besides the theoretically given differences in foot widths, it is necessary to respect also the toe length differences.

The aim of this study is to state, whether the dimensional proportionality of the feet of girls markedly changes with the age and whether it is necessary to respect these changes when designing the shoe lasts for the manufacture of children's footwear.

Methods

A random sample of 264 Mongolian girls ranging from 3 to17 years of age was divided into 4 groups according to their age (3-6, 7-11, 12-14 and 15-17). Foot characteristics were measured on each child's right foot using foot-measuring devices (a measuring tape); 2 dimensions: ball width and ball girth. Footprints were recorded from a standing full weight-bearing position. The footprints were analyzed and three dimensions were measured from each footprint: foot length, arch length and toe length. Starting from these data, matrices of the respective correlative coefficients have been computed. Among the given five variables, the following criterions have been taken into consideration: foot length (1), arch length (2), toe length (3), ball width (4) and joint girth (5). Correlative coefficients have been plotted with respect to the age of the girls.

Results

Following figures depicted the correlation results gained from the experiment. Fig. 1 visualizes the dependence of the correlative coefficient on the age of Mongolian girls. The correlative coefficient between the foot length and the arch length (1--2) decreases with the age, whereas the correlative coefficient between the arch length and the toe length (2--3) decreases more markedly in dependence on the age. This fact refers to an interesting conclusion with a practical effect. With the younger girls, the dimensional proportionality of their feet is more identical (the individual deviations from the average values are smaller) and this value decreases with the age. Above all, it is the toe length which changes. This statement would confirm the justification of the standardization of the shape of the lasts intended for the manufacture of the shoes for the smallest children (in our case, the girls).

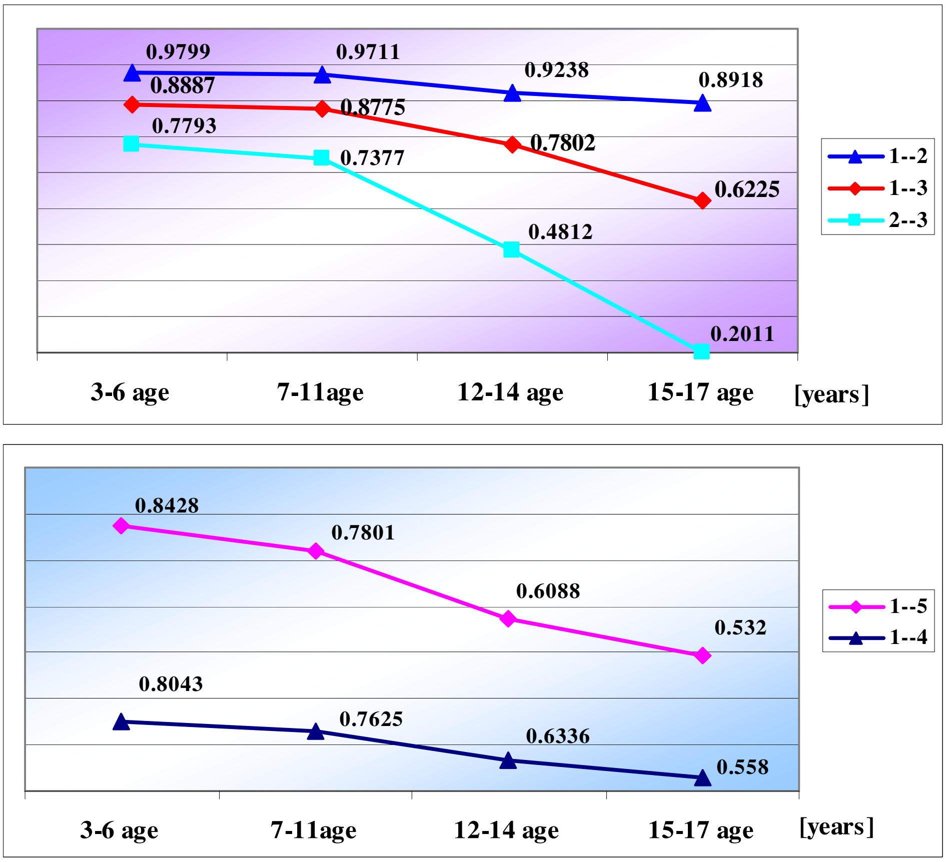

Fig. 1. The dependence of the correlative coefficients, by age group

Discussion

The given study has only been made on a group of girls and, for an objective confirmation of the existing dependence, there would be advisable to perform the same study with the boys and with the adult population as well. There is namely the fact evident, that the fashion designers very frequently do not respect the principles of the dimensional shapes of the lasts. Owing to the fact that most deformities of the children´s feet arise within the period, when their feet grow, it is necessary to pay more attention to this problem.

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The Influence of Heel Height on Patellofemoral Joint Stress during Walking

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Introduction

Patellofemoral pain (PFP) is a common condition of the lower extremity. It has been reported that women are more likely to experience PFP when compared to men.[1] Although wearing high heeled shoes has long been considered a predisposing factor with respect to the development of PFP in women, it is not known how wearing shoes of varying heel heights affects patellofemoral joint (PFJ) loading. The purpose of this investigation was to quantify the influence of heel height on PFJ forces and stress during walking.

Clinical Significance

Evaluating the influence of heel height on PFJ stress is important to better understand the relationship between footwear, joint loading and PFP. Information obtained from this study will be useful with respect to educating patients on proper footwear selection.

Methods

Eight healthy women participated in this study (mean age: 25.0 ± 3.1, height: 161.6 ± 5.4 cm, and weight: 55.4 ± 8.1 kg). Lower extremity kinematics (8-camera Vicon motion analysis system; 120 Hz), and kinetics (AMTI force plate; 1560 Hz) were obtained during at a fixed walking speed (82 m/min) under 3 different shoe conditions that varied in heel height (low: 1.27cm; medium: 6.35cm; high: 9.53cm). A previously described biomechanical model was used to estimate patellofemoral joint stress.[2] Briefly this model uses subject input variables (i.e. knee joint kinematics, net knee joint moment) as well as variables from the literature (i.e. knee moment arms, quadriceps force/patella ligament force ratios and joint contact area). Model output was PFJ reaction force, PFJ stress and utilized contact area as a function of the gait cycle. We also evaluated the stress-time integral among shoe conditions by comparing the area under the stress curve. Separate one-way ANOVA's with repeated measures were used to compare output variables among the 3 shoe conditions. For all ANOVA tests, post hoc comparisons consisting of paired t-tests were employed using a Bonferroni adjustment. All significance levels were set at $p < 0.05$.

Results

For all conditions tested, peak PFJ stress occurred during the weight acceptance phase of gait (Fig.1). The ANOVA comparing peak PFJ stress among the 3 shoe conditions was significant ($p < 0.001$). On average, significantly greater peak PFJ stress was observed with increasing heel height (Fig. 1). Similar results were found for the peak PFJ reaction force, utilized contact area at the time of peak PFJ stress, and the stress-time integral (Table 1).

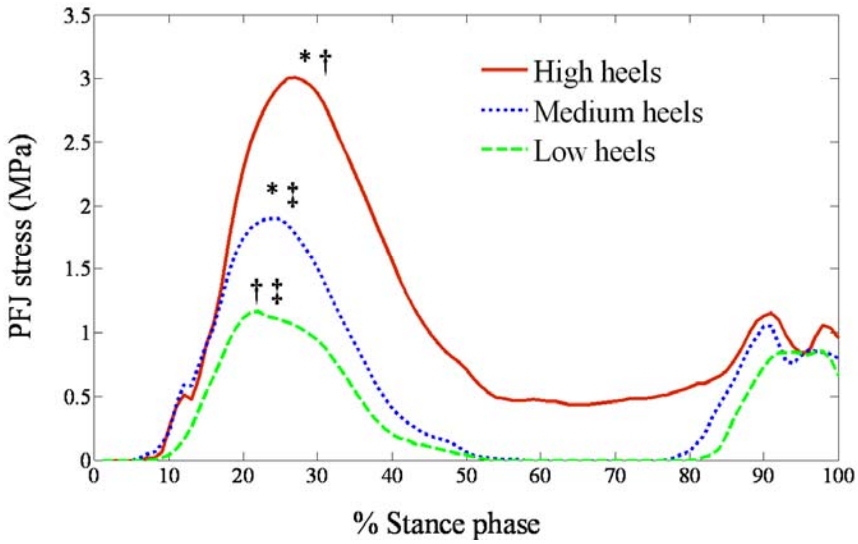

Fig.1. PFJ stress during stance phase for the 3 shoe conditions. All heel height comparisons were found to be statistically significant (p<0.016).

Table 1. Statistical results comparisons during the three shoe conditions.

	PFJ stress-time integral (MPa*% Stance phase)	Peak PFJ reaction force (N/kg)	Peak utilized contact area (mm^2)
Low heels	45.0±20.6†‡	6.3±2.8†‡	156.8±22.2†‡
Medium heels	65.8±34.3*‡	8.3±4.9*‡	175.8±22.9*‡
High heels	117.5±52.4*†	11.8±5.9*†	191.3±18.3*†

Values are presented as mean ± SD. All heel height comparisons were found to be statistically significant for each variable (p<0.016).

Discussion

Our results demonstrate that increasing heel height has a substantial influence on PFJ loading. In particular, we found an 89% and 160% increase in peak PFJ stress and the PFJ stress-time integral respectively between the low and high heel conditions. Although a 22% increase in peak utilized contact area was found, the significantly greater increase in peak PFJ reaction force (87%) contributed to the overall increase of peak PFJ stress during high heeled gait (Table 1). Our results reinforce clinical observations linking the wearing of high heeled shoes to PFP.

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COMPARISON OF GAIT CHARACTERISITCS FOR STANCE CONTROL KNEE-ANKLE-FOOT ORTHOSES USER

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Introduction

The number and variety of available stance control knee-ankle-foot orthoses (SCKAFO) has increased within the past few years making it vital to determine which will be more appropriate and effective not just for a diagnosis, but for a specific user.[1,2] For appropriate candidates, it offers either free knee motion or extension assist in swing phase and locking or knee flexion resistance in stance phase. The microprocessor SCKAFO offers loading support during extension through use of a wrap spring clutch, but the addition of a microprocessor and battery make the device more cumbersome.[1,2,3] Several papers compare kinetic and kinematic variables with adaptation over time of one single orthosis in experienced vs. novice users[2] or comparing locked knee vs. dynamic stance control. [1,3] This study examines the gait characteristics of one patient with a spinal cord injury wearing a powered dynamic knee versus his daily use pneumatic extension assist SCKAFO.

Statement of Clinical Significance

The evolution of the SCKAFO claims to allow greater freedom through more symmetric and energy efficient gait; to date, there have not been any studies comparing SCKAFOs. The patient's goals are to achieve instantaneous variation of velocity (i.e. to cross the street) descend stairs, and ascend hills reciprocally. His lower left extremity (LLE) has impaired strength throughout. His daily use SCKAFO offers effective ambulation, but the microprocessor SCKAFO might allow him to better achieve his goals. This paper will focus on the comparison of joint kinematics between the patient's two SCKAFO to determine the advantages and limitations of both.

Methods

The patient is a 38 year old male who sustained a L2-3 root transections resulting in monoparesis of the LLE in April 2003. His gait was evaluated in everyday footwear with his daily use SCKAFO (GX-Knee, Becker Orthopedic) and with a microprocessor controlled SCKAFO (Sensor Walk Orthosis, Otto Bock HealthCare LP) at a self-selected velocity. Gait was collected using a modified Cleveland clinic marker set, 6 force plates (AMTI, Watertown, MA) and a Vicon motion capture system; were analyzed using a custom BodyBuilder model and Polygon (Vicon Motion Systems Ltd. Oxford, UK).

Results

This report will compare the Becker to the Otto Bock (OB) brace. The Becker condition exhibits less pelvic elevation and greater trunk lateral flexion than the OB, with slightly increased L lateral trunk flexion at terminal stance in concert with increased hip hiking in preparation for L swing clearance. L hip

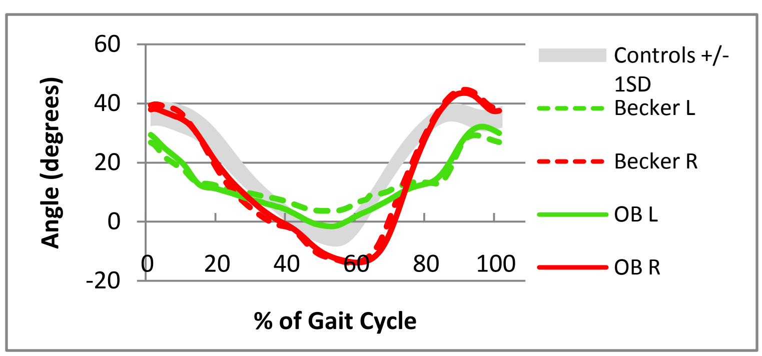

Figure 1: Average Hip Flexion/Extension of both SCKAFOs over gait cycle

flexion excursion is markedly decreased in both braced conditions, peak extension excursion is also decreased, although OB achieves extension (Figure 1). As seen in Figure 2, the L knee does not demonstrate loading response with either brace. However, there is a gradual, progressive knee flexion into swing in the Becker. OB L knee flexion is negative at heel strike through terminal stance with a lower peak flexion than the Becker. The L knee reaches early terminal extension in both conditions prior to initial contact. Stance R plantar flexion (pf)/dorsi flexion (df) excursion is closer to controls in the Becker; peak df in the OB condition is 20°, 22° in Becker. Both braces demonstrate a lift off of the foot during swing and increased slope of pf at loading consistent with footslap (more pronounced in Becker). Both conditions show R ankle pattern is consistent with vaulting, with OB more pronounced than Becker.

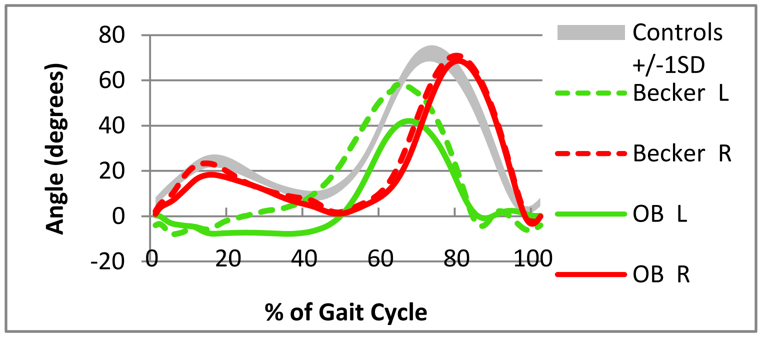

Figure 2: Average Knee Flexion Extension of both SCKAFOs over gait cycle

Discussion

Previous studies have found increased knee flexion and alterations in hip, pelvis, and vaulting compensatory motions with the use of the SCFKAO compared to locked knee.[1,2,3] In this case, the weight of the SCKFAO (Becker 1.3 kg, OB 3.2 kg) may have contributed to the higher peak L hip extension achieved in the OB vs. Becker as opposed to muscle activation (Becker -4°, OB 2°). Loss of LLE strength and muscle mass leading to lack of tissue stabilization, along with the weight of the OB SCKAFO, may have caused posterior bone movement leading to the apparent hyperextension (L knee flexion of -8°) of the left knee throughout terminal stance. The R ankle pattern consistence with vaulting may also be attributed to the weight of the KAFO. After additional gait training and modification to the OB SCKFAO, the patients gait will be re-evaluated in both braces.

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Neuromechanical and Neurophysiological Examination of Walking with an Ankle Foot Orthosis in Individuals with Incomplete Spinal Cord Injury

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Introduction

After an incomplete spinal cord injury (ISCI), a compensatory orthotic device such as an ankle foot orthosis (AFO) is commonly used to stabilize the ankle joint and aid toe clearance during stepping. Compensatory stepping achieved with an AFO has led to the assumption that such devices could be integrated in newer, neurobiologically driven, recovery-based interventions such as locomotor training (LT) for individuals with ISCI. In spite of the appeal of such compensatory strategies, their use in recovery-based interventions such as locomotor training for individuals with ISCI is still controversial.[1] This is due to the lack of information about the use of the device in optimizing or hindering afferent input from joint, muscle and cutaneous receptors fundamental to the training.[2] After ISCI, pattern generating neural networks within the spinal cord increase their reliance on motion-related afferent input from these receptors for maintaining locomotor control.[3, 4] Evidence from previous studies indicate that terminal hip extension and unloading are critical sensory inputs required for the afferent initiation of the transition from stance-to-swing during stepping.[2, 5] Limiting hip extension and/or transfer of weight on either limb delays the transition from stance-to-swing and swing-to-stance and limits forward progression of the body. Limiting ankle excursion with an AFO may alter the interconnected knee and hip joint motion and in turn negatively influence the afferent information critical for stepping.

Clinical Significance

Assessment of such devices in recovery-based training paradigms such as locomotor training will provide information about the degree of conformity of such clinical strategies with the principles of neurobiological control of walking. Accordingly, in this study, mechanics of stepping and soleus H-reflex modulation with the AFO will be evaluated to characterize the motion-related afferent input being processed at the level of the spinal cord. Evidence of altered/improved stepping mechanics and reflex modulation would reflect the failure/success of compensatory devices to provide task-specific sensory input necessary for walking recovery. Knowledge of results from this study would influence clinical decision-making for the use of such devices in recovery-based training interventions.

Methods

Nine persons having ISCI classified as ASIA D, 10-253 months post-injury, mean age 45 ± 13 years participated in the study. Each participant provided informed consent prior to participation in the study. Participants were instrumented with light reflective markers and walked on a split-belt instrumented treadmill with and without an AFO at self-selected speed. Motion data and ground reaction forces (GRFs) for each leg were captured during 30 seconds trials. All participants wore a minimally restrictive, posterior leaf spring AFO (PAFO) on their more impaired limb. Soleus H-reflexes were elicited in the mid-stance and mid-swing phase while walking with and without the PAFO. Soleus H-reflexes were also elicited in standing which served as control reference. Kinematic and kinetic data for all subjects were normalized to the gait cycle, averaged and compared using a Hotelling's T^2 test statistic. The H-reflex amplitude was normalized to the maximum m-wave amplitude observed in each phase (H/M ratio) and compared between conditions with a paired T-test.

Results

Neuromechanically, the use of a PAFO significantly decreased hip extension in terminal stance (without PAFO -2.6± 10.6° and with PAFO -1.2± 9.7°, p=0.02). Neurophysiologically, a significant increase (without PAFO: 0.13±0.10 & with PAFO 0.29±0.14, p=0.004) in soleus H-reflex amplitude was observed in mid-swing.

Discussion

In our study, the use of a minimally restrictive PAFO decreased hip extension in participants with ISCI. The observed decrease could impact the provision of at least one critical afferent input key to the restoration of walking. Furthermore, use of more rigid devices is likely to exaggerate our findings. Consequently, if the goal of recovery based interventions such as locomotor training is to provide optimal limb kinematics, the use of a PAFO for stepping would not coincide with the principles of training. Also, in the absence of a change in the ankle-foot orientation or stretch at the ankle joint with a PAFO, the increase in soleus H-reflex amplitude is suggestive of an increase in afferent inflow in the mid-swing phase of walking. However, increase in afferent input in the mid-swing phase of walking might not be favorable for retraining walking in the presence of impaired reflex modulation as observed after ISCI. Therefore, our study demonstrates that the integration of clinically acknowledged stepping aids such as the PAFO's during locomotor training could be counterproductive to the recovery of walking post SCI. Therefore such devices should be chosen only after careful consideration of outcome for training.

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Acknowledgements

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MILITARY SERVICE MEMBERS WITH BLAST-RELATED TRANSTIBIAL AMPUTATIONS: TEMPORAL-SPATIAL ASPECTS OF WALKING WHILE CARRYING MILITARY LOADS

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INTRODUCTION

Approximately 660 U.S. military personnel have sustained lower extremity (LE) amputations while serving in Afghanistan or Iraq of which 295 have experienced unilateral transtibial amputations (TTAs). Many of these service members are highly motivated to remain in service and return to theater. To actively participate in ground combat duties, amputees must meet fitness and functional standards including marching with rucksack loads.

Data on carrying of rucksack loads by amputees is lacking. However, biomechanical research on the carrying of loads by uninjured individuals indicates that changes occur in gait and posture that likely serve to maintain stability and mitigate forces as the external loads on the body increase[1]. A change reported in investigations conducted at fixed walking speeds is an increase in the double support period of the gait cycle with increased loads[1, 2].

To address knowledge gaps in load carrying by amputees, we compared the biomechanical responses of service members who incurred a unilateral TTA with uninjured soldiers. We hypothesized that individuals with TTAs would evidence shorter step lengths and double support periods when carrying loads than individuals with intact limbs.

CLINICAL SIGNIFICANCE

Findings from the study will help the rehabilitation community to better understand the nature and extent of the physical demands placed on LE amputees and on their prostheses. Outcomes may aid in development of rehabilitation strategies and prostheses for improving the readiness of military amputees to safely return to active duty.

METHODS

Temporal and spatial gait variables were calculated from data recorded as four service members with TTAs (Means: 29.3 yrs; 178.6 m; 92.8 kg) and four control subjects (Means: 20.5 yrs; 179.3 m; 81.2 kg) walked on a treadmill at two speeds (1.34 or 1.52 m/s) under three load conditions. These data are part of a larger study evaluating biomechanical and metabolic responses of 12 males with TTAs and 12 uninjured males. Each walked for 10 min with no load, a rifleman's fighting load (vest, 21.8 kg), and a rifleman's approach march load (ruck,32.7 kg). Motion capture data were collected for three 10 sec trials once steady state walking was achieved. The trials were processed and temporal-spatial outcome measures were calculated using Visual 3D (C-Motion Inc, Rockville, MD).

RESULTS

The double support period increased with load weight, and decreased with speed (Figure 1). The amputees displayed shorter step lengths (Figure 2) than the controls under all test conditions and took longer steps on their prosthetic LE. Figure 3 illustrates the asymmetries in single limb support time of the amputees; spending more time on their intact LE. Step width appears to be similar between amputee and control subjects (Figure 4).

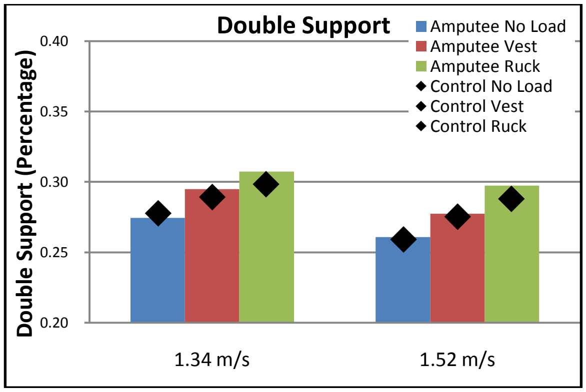

Figure 1

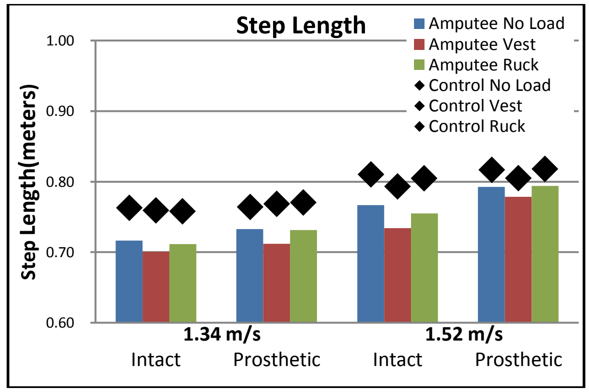

Figure 2

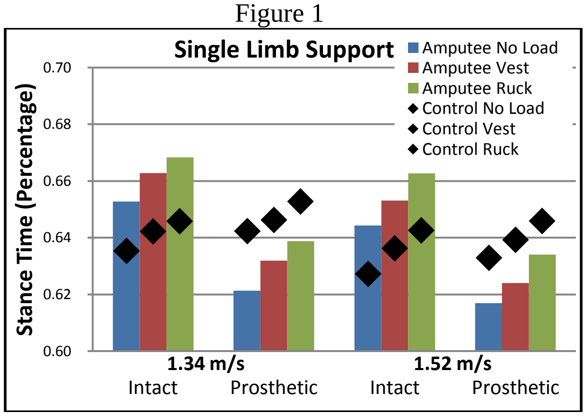

Figure 3

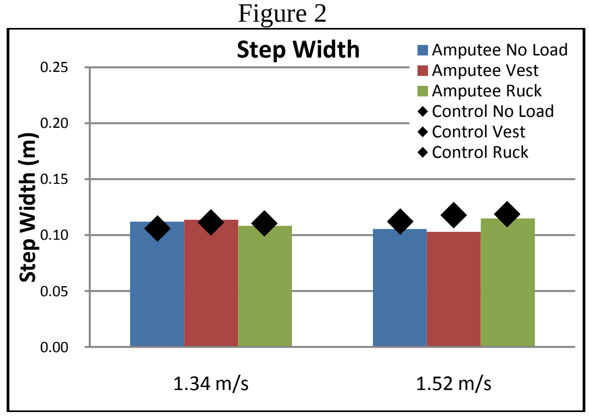

Figure 4

DISCUSSION

Findings regarding effects of walking speed and load mass on double support are consistent with the literature[1, 2]. The amputees and the controls showed similar trends for changes in gait variables as a function of load. However, the asymmetric and shorter step lengths of the amputees under all test conditions suggest that they may be less stable when carrying rucksack loads than uninjured individuals and also more likely to incur overuse injuries during prolonged marching with military loads. We hypothesize that the step width data is significantly influenced by the constraints of the treadmill and may not reflect what occurs in over ground walking. Completion of the study will determine whether initial trends in the data are statistically significant.

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ACKNOWLEDGEMENTS: We would like to thank the Military Amputee Research Program for support of this research.

Disclaimer: *the views expressed in this abstract are those of the authors and do not reflect the official policy of the Department of Army, Department of Defense, or U.S. Government.*

Advanced Technologies in the Development of Dynamic Pedorthosis

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Introduction: The Denis-Browne splint and its modifications have been used for decades to prevent clubfoot relapse or as night splints following Ponseti's treatment. The technique to make a traditional orthosis is dependent upon skilled professional judgment using non-weight bearing casting or foot imprints. The purposes of this research were 1) to use dynamic plantar pressure data, along with CT scan data to develop pedorthotics for the treatment of clubfoot deformity, 2) develop a computer aided design (CAD) and Finite element Analysis (FEA) that will predict the required orthotic to correct the deformity.

Clinical Significance: This dynamic pedorthosis should assure an appropriate design and improve consistency of technique for pedorthotics.

Methods: Five normal children with average age of 7.2-year-old (2 girls and 3 boys) and five clubfoot patients with average of 6-year-old (1 girl and 4 boys) were recruited to obtain dynamic plantar pressure using EMED Pressure System, and three-dimensional geometry from a CT scan. Four patients were diagnosed as bilateral clubfoot deformities and one as unilateral clubfoot. Treatment history included a posterior medial release for two patients, series casting for two patients, and a complete subtalar release for one patient. Functional outcomes were reported using Foot and Ankle Questionnaires, Pediatric-Parent Questionnaires as well as clubfoot-specific instruments.

A CAD model of a pedorthosis was constructed, containing polymer and foam elements (PTC, Needham, MA). The FEA was applied to analyze the peak pressure changes (MSC, Santa Ana, CA). The pedorthotic was constructed using rapid prototyping (RP) technologies.

Results: Table 1 shows an increased contact area at the mid-foot, mid-forefoot and reduction in the contact area at the hallux compared to the normal foot. However, Table 2 shows increased peak pressure on the lateral forefoot.

Table 1 Significant differences in the contact area between normal and children with clubfoot ($p < 0.05$, SPSS).

Contact Area	Normal (Mean ± SD)	Clubfoot (Mean + SD)	Probability
Mid-Foot	10.8 ± 2.3	15.4 ± 6.4	0.02
Mid-Forefoot	10.8 ± 1.4	11.1 ± 3.9	0.002
Hallux	6.4 ± 1.3	4.2 ± 2.9	0.001

Table 2 Significant differences in the pressure between normal and children with clubfoot ($p < 0.05$, SPSS).

Peak Pressure	Normal (Mean ± SD)	Clubfoot (Mean + SD)	Probability
Lateral Heel	27.4 ± 4.3	18 ± 9.4	0.03
Lateral Forefoot	13.1 ± 3.8	17.4 ± 13.9	0.03
Mid-Forefoot	20.3 ± 3.4	27.7 ± 15.2	0.02

Five orthotics were developed using the CAD models. The computer simulation incorporated size and geometry of the various arches in the foot (Figure 1).

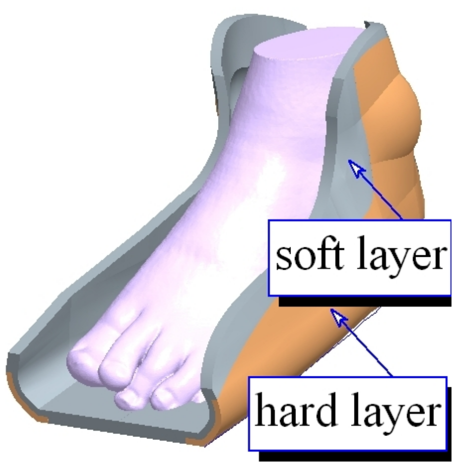

Figure 1: CAD model of an orthotic. A model generated from the CT data of the foot has been added to shown the relative size

Discussion: From Table 2, we are able to determine the deviation of the pressure matrix between the normal and the clubfoot, which is needed in the construction of the CAD and FEA models. Additionally, a CAD designed foam wedge was added to redistribute the plantar pressure, which reduced deviation of plantar pressure trajectory from 10% to 22% of normal. The short-term evaluation was documented that the patient was able to comply the compliance moderately.

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HOW DO ERRORS PROPAGATE IN HELEN HAYES AND 6 DoF MODELS OF GAIT?

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INTRODUCTION: The Helen Hayes model (HH) is a common model for gait data collection and reduction[1] and requires accurately placed markers on bony landmarks to calculate the locations of joint centers. One limitation of HH is that segments are not independent and that errors propagate through the segments. In contrast, a six degree-of-freedom model (6DoF) does not constrain the joints and localizes the errors to each segment. The purpose of this abstract is to compare HH and 6DoF during an identical single stride and to uncover sources of error in each model. In addition, we will compare how each model responds to a perturbation of pelvis segment orientation. Previous studies have shown that HH models were similar despite differences in filtering techniques[2]. Furthermore, these studies did not detail if the joint centers were aligned, how they were calculated or whether a knee alignment device (KAD) was used[2,3]. This abstract will report differences between HH and 6DoF, and how a perturbation affects the differences within the same model with identical filtering.

CLINICAL SIGNIFICANCE: The application of accurate models is essential if results are to be used to drive treatment planning and assess outcomes of interventions.

METHODS: A Vicon 612 motion capture system (Vicon Motion systems, Lake Forest, CA) was used to capture the 3D lower extremity gait kinematics and barefoot stance phase kinetics (AMTI, Watertown, MA) of a single volunteer (1.57 m, 51.3 kg). A single static calibration and dynamic walking trial (1.15 m/s) were taken using a hybrid marker set that allowed simultaneous collection of both models. Kinematic data was collected at 120 Hz and kinetic data at 360 Hz. Two HH models were created based on the same static calibration, one in Vicon and the other in V3D (C-Motion, Inc., Gaithersburg, MD). The 6DoF model was created in V3D. The dynamic trial filtered the marker data using a bidirectional Butterworth filter with a cutoff frequency of 6 Hz, and the ground reaction force data was filtered using the same filter with a cutoff at 25 Hz. The joint centers and segment coordinate axes were modified in the 6DoF model to be identical to the V3D HH model. Copies of the c3d files were imported back into Workstation to calculate the gait analysis variables, while identical files were processed in V3D. This ensured that all models were filtered identically, and that any differences could be attributed to the models. Perturbations were applied to the models in V3D by applying a 2 cm offset in the X, Y, and Z lab coordinate system to the SACRUM and the R ASIS targets tilting (frontal plane) and rotating (transverse plane) the pelvis segment. After processing, the moment data were exported within a single step and the angle data were exported between a single stride. Hip, knee, and ankle angles and moments were compared between models.

RESULTS: The joint centers in the HH were identical to 6DoF in V3D; however differences existed between the implementation of the HH in V3D and Vicon. Originally, the maximum joint differences were at the knee (9+ mm in the ML direction). However, a correction was made to the Vicon HH model to account for a 9mm diameter markers on the KAD. The KAD is designed for use with 25 mm markers. After the correction was made, maximum geometric difference was at the hip joint, R=2.18mm and L=2.70mm).

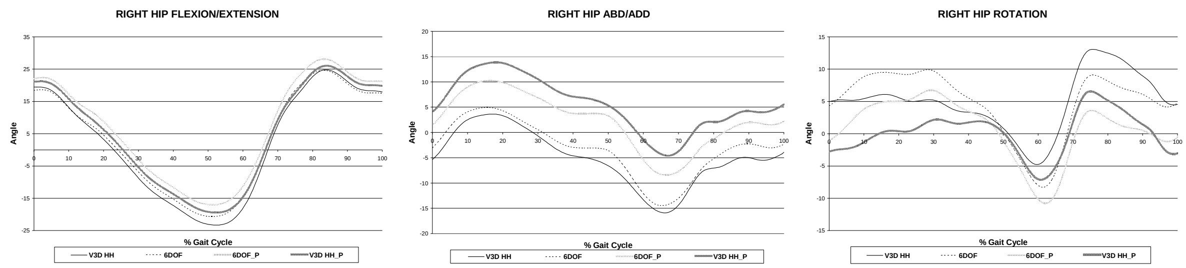

Figure 1: Hip angle graphs for HH, 6DoF, HH perturbed, and 6DoF perturbed models

The pelvis perturbation caused errors for both models at the hip, particularly in hip ABD/ADD with the HH model showing larger error (10.5° vs. 5.5°; 0.23 Nm/kg vs. 0.07 Nm/kg). The angles and moments at the knee and ankle were not affected by the perturbation in the 6DoF model. The HH model had small differences at the ankle (up to 0.2° and 0.001 Nm/kg), but larger errors (up to 4.9° and 0.009 Nm/kg) occurred at the knee.

Table 1: Kinematic and Kinetic Differences between Models

	Angle Differences in Degrees				Moment Differences scaled by Bodyweight (51.3 KG)		
	HH v 6DoF	HH Perturbed	6DoF Perturbed		HH v 6DoF	HH Perturbed	6DoF Perturbed
R Hip Angle (x)	1.51	2.60	3.64	R Hip Moment (x)	0.0562	0.0195	0.0315
R Hip Angle (y)	1.76	10.53	5.50	R Hip Moment (y)	0.0240	0.2273	0.0238
R Hip Angle (z)	2.94	4.81	3.94	R Hip Moment (z)	0.0065	0.1079	0.0711
R Knee Angle (x)	2.08	1.05	0.00	R Knee Moment (x)	0.0251	0.0094	0.0000
R Knee Angle (y)	2.09	4.88	0.00	R Knee Moment (y)	0.0094	0.0051	0.0000
R Knee Angle (z)	3.04	1.42	0.00	R Knee Moment (z)	0.0085	0.0167	0.0000
R Ankle Angle (x)	0.66	0.10	0.00	R Ankle Moment (x)	0.0080	0.0002	0.0000
R Ankle Angle (y)	2.69	0.15	0.00	R Ankle Moment (y)	0.0139	0.0010	0.0000
R Ankle Angle (z)	2.37	0.21	0.00	R Ankle Moment (z)	0.0081	0.0004	0.0000

DISCUSSION: The joint centers calculated by the HH model in Vicon and V3D were comparable. A brief comparison between Vicon and V3D showed almost identical data in the sagittal plane with increasing divergence in the frontal and transverse planes. A detailed description of these differences is outside the scope of this abstract. The major kinematic difference between the HH and 6DoF occurred at the knee (3.04°) most likely due to soft tissue movement of the lateral knee marker. The kinetic differences between HH and 6DoF were less than 6% of bodyweight. The advantage of 6DoF is that kinematic and kinetic errors from the perturbation did not propagate to the knee and ankle (Table 1). Our results indicate that labs using a KAD other than 25 mm markers must make additional offsets either in the model or by physical modification of the KAD. Failure to do so will result erroneous knee and ankle joint centers.

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Inter- and Intra-Observer Variability of Marker Placement. David J. Gutekunst, Melanie Koleini, Dequan Zou, Robert H. Deusinger. Human Biodynamics Laboratory, Program in Physical Therapy, Washington University in St. Louis

Introduction: The primary goal of quantitative movement analysis is to describe mechanical aspects of the musculoskeletal system during movement tasks (1). In analyses using high-speed video to track markers placed on the skin, accuracy and reliability of movement data can be significantly reduced by soft tissue artifact and by marker placement error (2; 3).

Clinical Significance: Attempts have been made to assess optimal marker attachment methods (2), minimize inter-marker pair distances during movement (4), and quantify typical errors associated with quantitative gait analysis (5). The purpose of the present study was to quantify differences between repeated marker placements, both in terms of inter-observer difference and the within-session and between-session intra-observer differences.

Methods: Eighteen healthy participants (10 female) volunteered and signed an informed consent document approved by the university IRB. Data were collected by two observers on two sessions separated by 24-48 hours, for a total of 4 applications to obtain inter-observer difference and 2 forms of intra-observer difference (within- and between-session).

After applying a set of 28 spherical (14 mm) retro-reflective markers bilaterally on the pelvis, thigh, shank, and foot using double-sided toupee tape, Observer 1 used an invisible ink pen to circle the plastic base of each marker. Next, Observer 1 removed the markers and used a UV pen to illuminate the invisible ink circle, then placed an invisible ink dot in the center using a custom-made stencil. Observer 2 then repeated the entire process, except a Crayola® washable ink pen was used. The distance from the invisible ink center dot to the Crayola ink center dot was measured using a high-precision (0.01 mm) digital caliper. The center-to-center distance served as the inter-observer difference for each location. The third marker application of Session 1, placed by Observer 1, was marked with Crayola ink to allow for measurement of the within-session, intra-observer difference.

Session 2 occurred 24-48 hours after Session 1 and consisted of Observer 1 placing the entire marker set to allow for measurement of between-session, intra-observer differences by comparing the Session 2 center dots to the invisible ink dots made by Observer 1 during Session 1. In total, the 4 complete marker applications allowed for the following comparisons for each subject: within-session, inter-observer reliability; within-session, intra-observer reliability; and between-session, intra-observer reliability. The average difference and standard deviation of the difference between marker applications was computed at each of the target locations on the pelvis and the right side of the lower extremity.

Results: The subject cohort (n=18, 10 females) was 24.6 ± 2.1 yrs, 169.9 ± 11.5 cm, 65.7 ± 11.1 kg, and had BMI of 22.6 ± 1.8 kg/m^2. Within-session, inter-observer difference was larger than either of the intra-observer differences at all individual anatomical locations (Table 1). When intra-observer measurements were averaged across all locations, the within-session and between-session differences were roughly comparable. Bony landmarks on the shank and foot (tibial tuberosity, malleoli, metatarsal heads) had a lower average difference than anatomical locations on the pelvis and femur.

Table 1. Average (± SD) difference (mm) by anatomical location.

Location	Within-session Inter-Observer	Within-session Intra-Observer	Between-session Intra-Observer
ASIS	11.08 (10.30)	8.89 (9.87)	9.56 (10.79)
Iliac Crest	15.92 (8.42)	10.81 (5.90)	12.28 (5.85)
PSIS	12.21 (10.02)	7.02 (5.04)	7.55 (4.35)
Greater Trochanter	16.99 (8.14)	9.42 (6.76)	16.69 (15.42)
Med Femoral Epicondyle	13.30 (7.10)	10.49 (9.21)	8.85 (5.21)
Lat Femoral Epicondyle	15.01 (6.28)	10.66 (7.52)	12.59 (6.37)
Fibular Head	11.49 (8.84)	9.07 (5.77)	8.60 (5.35)
Tibial Tuberosity	7.46 (4.72)	5.70 (2.85)	5.43 (2.73)
Lateral Malleolus	4.03 (3.13)	3.02 (2.02)	3.24 (2.12)
Medial Malleolus	5.26 (2.18)	3.82 (2.44)	3.09 (1.36)
Calcaneus - AT insertion	11.24 (7.68)	5.06 (4.56)	4.25 (2.69)
2nd metatarsal	4.24 (2.56)	3.40 (1.76)	3.43 (1.20)
5th metatarsal	3.28 (2.34)	2.72 (1.26)	3.24 (2.05)
Average across Locations	***10.73***	***7.27***	***8.10***

Discussion: These results suggest that a single observer is more reliable in placing anatomical markers than multiple observers and that easily-palpable bony landmarks are more reliably located than soft-tissue landmarks (5). The results also suggest that within- and between-session variability are roughly equal for a single observer, although the somewhat small difference we observed could be due to the relatively short period (24-48 hrs) between sessions. A key finding of this study is that differences in marker placement are much larger than typical motion capture measurement error or soft tissue artifact (4). Thus, to improve accuracy and reliably of movement analysis, concerted effort should be directed towards improving the reliability and accuracy of marker placement.

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Skin motion artifact in the forearm and wrist: A comparison of surface markers and CT image based bone registration in a cadaver model.

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Introduction

Skin surface mounted markers (SSM) or other skin mounted sensors are frequently used to measure 3-dimensional joint kinematics. SSM is relatively simple to implement and allows measurement of kinematics in a non invasive manner. Disadvantages of SSM methods are uncertainty in locating landmarks on the bones whose motions are being tracked and skin motion artifact affecting the marker locations relative to the bones throughout dynamic activities. CT based bone registration (CTBR) can directly measure the location of bones in 3 dimensions. CTBR has the limitations that it is typically constrained to static positions and requires the use of ionizing radiation, excluding CTBR from use in dynamic studies of wrist motion. The purpose of this study was to examine the use of SSM compared to CTBR while tracking wrist motion in a cadaver model, with CTBR being considered as the "gold standard".

Clinical Significance

It is important to document and minimize potential error due to skin motion artifact associated with surface mounted markers. The choice of location of markers and clusters should be based upon quantification of the markers' motions relative to bone.

Methods

An external fixator was attached to the third metacarpal and radius of a cadaver forearm and used to fix the wrist in 10 static positions: neutral, maximum flexion, maximum extension, maximum radial deviation, maximum ulnar deviation and 5 positions along a "dart throwing motion" (DTM) path. A SSM system and CTBR were simultaneously used to measure the wrist position at each of the 10 positions. 11 markers were placed as the SSM system to track the motion of the 3rd metacarpal and radius. The entire hand and distal half of the radius was CT scanned (voxel resolution .49 mm^3) with the markers in place. The surface markers, radius and 3rd metacarpal were segmented from the CT image and 3-D models were generated with Mimics 9.11 (Materialize, Ann Arbor, MI). Wrist kinematics were then generated from both the SSM positions and CTBR positions.

For SSM, wrist kinematics were calculated by grouping the surface markers into two marker clusters corresponding to the radius and 3rd metacarpal. The radial cluster utilized markers placed over the radial styloid (RDST), Lister's tubercle (RDRD) and a rigid three marker triad placed over the posterior aspect of the radius (RDT1,2,3). The 3rd metacarpal cluster used markers over the distal end of the bone at both the dorsal (3MHD) and volar (3MHV) sides, a maker over the proximal end of the bone at the dorsal side (3MBD), and a rigid three marker triad (MCT1,2,3) placed over the center of the bone. For CTBR, motion of the radius and 3rd metacarpal was determined using previously established bone registration techniques[1]. Wrist kinematics (flexion-extension and radial-ulnar deviation) were calculated with respect to a radial coordinate system[2] using the long axis of the 3rd metacarpal[3] using Euler angles for both techniques.

Results

The relative marker movements with respect to their associated bones ranged from .17 mm to 5.78 mm throughout the positions tested (Fig. 1). The greatest movements were associated with the RDST marker for the radius, and the 3MHV marker for the 3rd metacarpal. The mean angular difference between the SSM and CDBT kinematic measurements across all positions tested was 5.6° ± 4.4° for flexion/extension and 5.0° ± 2.7° for radial/ulnar deviation (Fig. 2).

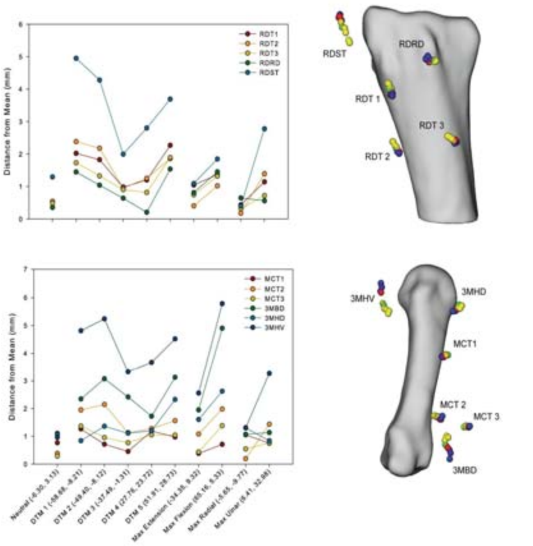

Figure 1. Relative marker locations across wrist positions.

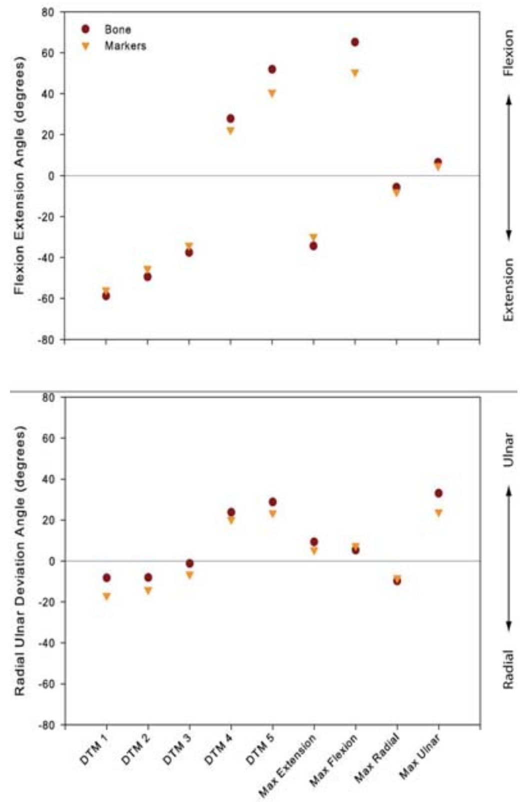

Figure 2. Angular positions as measured by bone based and marker based systems.

Discussion

This study provides an estimate of error due to skin motion artifact for SSM tracking of wrist motion, but is limited by the use of a cadaver model in only one specimen and static positions. At the radius, relative marker movement was greatest for the RDST marker and was much lower for the RDRD marker. This suggests that for tracking of the radius, a marker array over the radial styloid should be avoided. For the 3rd metacarpal, the triad placed over the center of the bone worked reasonably well. Accuracy could be improved by tracking each segment with the 3 markers which exhibited the least skin motion.

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Repeatability of Knee Joint Rotations for Eleven Clinically Relevant Motor Tasks

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Introduction: A custom gait analysis protocol specifically focused on knee kinematics and kinetics potentially can advance clinical evaluation of the knee. While developing such a custom protocol, the present study measured the repeatability of knee kinematics from a standard whole body gait analysis model (Plug-in Gait [PiG] with Knee Alignment Device[1] [KAD]) during 11 clinically relevant motor tasks. Previous studies measured repeatability of knee rotations for normal walking, when using the model on which PiG is based[2], and also when using the PiG-KAD[1] protocol. During gait, such standard models facilitate whole body analysis rather than focused analysis of the knee because of the small out-of-sagittal plane knee motion[1,2] and large skin motion artifacts associated with the thigh. A protocol that includes a variety of clinically meaningful motor tasks instead may better address study of the knee, if the motion kinematics are reasonably repeatable and have larger ROMs than in gait.

Clinical Significance: *In vivo* knee kinematics can be studied within a longitudinal follow-up and compared before and after surgical interventions. If the motor tasks studied here have acceptable within- and between-day repeatability and sufficiently discriminate between subject groups, then they may be useful for more detailed clinical studies of the knee.

Methods: Four healthy male volunteers (mean age = 35 ± 10 years, body mass = 82 ± 30 kg) each underwent three gait analysis sessions on different days, managed by a well-trained physical therapist, using the same protocol[1] with an 8-camera Vicon system and 2 AMTI forceplates. Subjects performed a set of 11 motor tasks per session (Table 1). One trial per motor task, per session, was used in this pilot study. Pivoting tasks required the subject to move forward and then change direction by 90°, with the support leg rotating either internally or externally. Kinematics were taken for the pivoting knee only. For the lunge, data were taken for the deep flexing knee over forceplate contact. All data were normalized in time according to a 100% cycle. Endpoints of cycles were detected with forceplates, except for chair rise and squat, whose cycle endpoints were detected visually on the curves. Intra- and inter-subject repeatability values were calculated across the subjects. The repeatability parameters are the mean absolute variability (MAV) and mean relative variability (MRV%) (Ferrari et al[3]), which give worst-case estimates of repeatability when limited trials are available. The MAV takes the range at each %cycle among all curves of interest, then averages these ranges. The MRV% divides the MAV by the full range of the average curve.

Results: The average intra- and inter-subject repeatability values for the 3 knee rotations and 11 tasks are summarized in Table 1. Within a subject, walking had the smallest flexion/extension (F/E) MAV, whereas walking with an internal pivot had the smallest ab/adduction (Ab/Ad) and in/external rotation (I/E) MAVs.

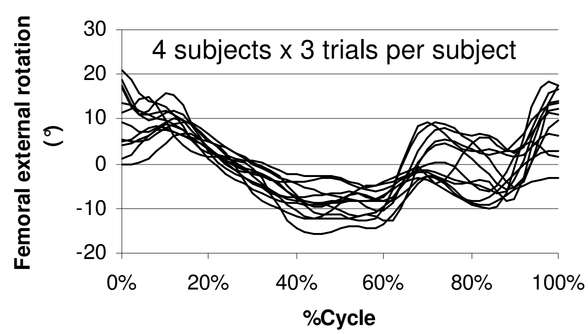
Discussion: Based on the MAVs, walking showed the most repeatable F/E measurements, while walking with an internal pivot (Figure 1) showed the most repeatable Ab/Ad and I/E measurements. Future studies that focus on repeatable knee Ab/Ad and I/E can consider analyzing this motor task. Walking repeatability values were within a few degrees of previous studies that analyzed fewer subjects and tasks[1,3]. The small differences may be due to different applications of the protocol, but as the results indicate they are mainly introduced by the different subjects analyzed. Inter-subject variability was several times greater than the mean intra-subject variability for all measurements. This suggests that the protocol analyzed has sufficient repeatability to distinguish subjects performing a variety of tasks. Repeatability was better for Ab/Ad than for I/E overall, which suggests that improved knee models should focus on I/E rotation. It should be noted that the most repeatable measurements are not necessarily the most accurate.

TABLE 1*

	Intra-subject repeatability						Inter-subject repeatability					
	MAV			MRV%			MAV			MRV%		
Motor Task	F/E	Ab/A	I/E	F/E	Ab/A	I/E	F/E	Ab/A	I/E	F/E	Ab/A	I/E
Walking	**5.5**	**2.8**	**7.1**	**9%**	**32%**	**56%**	**13.1**	**8.8**	**18.7**	**22%**	**117%**	**145%**
Walking with external pivot	7.6	2.8	***6.6***	13%	***31%***	***37%***	18.3	9.4	***17.9***	31%	139%	116%
Walking with internal pivot	6.1	***2.3***	***4.8***	10%	33%	***21%***	21.7	10.7	***12.0***	36%	200%	***62%***
Stair ascent	6.9	4.8	8.4	**8%**	39%	66%	19.6	16.8	20.6	23%	164%	197%
Stair descent	8.0	3.3	8.2	13%	40%	***56%***	18.4	10.1	21.2	32%	138%	174%
Stair descent with external pivot	7.8	3.2	9.0	13%	35%	***43%***	16.7	9.8	22.0	28%	141%	122%
Stair descent with internal pivot	8.6	3.2	7.7	16%	36%	***33%***	22.4	13.5	***16.2***	43%	174%	***70%***
Lunge	12.9	5.1	7.3	17%	57%	***38%***	32.8	19.7	***17.0***	42%	366%	***103%***
Chair rise	8.4	3.3	8.5	10%	51%	***56%***	22.4	13.7	20.3	27%	358%	126%
Squat, mild	13.4	3.9	7.9	16%	107%	79%	50.4	15.4	27.0	64%	488%	354%
Squat, deep	15.9	7.6	10.9	16%	66%	66%	54.7	26.5	26.0	53%	319%	149%

** Values are italicized that show repeatability equal to or better than normal walking.*

Figure 1. Knee rotation (+EX, -IN)



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A Femoral Epicondylar Frame for accurate transversal kinematics of the femur
A validation study using fluorscopy

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INTRODUCTION

Optoelectronic recording of skin mounted markers is used to assess knee kinematics in daily tasks. However, skin artefacts will hamper accurate measurements. For the shank these errors are limited [1] but especially the transversal plane errors can be considerable for the thigh [2]. Recently, we designed the Femoral Epicondylar Frame (FEF), inspired by a device presented by Houck et al. [3], using increased clamping pressure on custom made epicondylar attachments. This study was conducted to evaluate the precision and accuracy of the FEF in the patient target population using fluoroscopic stereometry of an knee prosthesis, which is accurate within 0.3° [4].

CLINICAL SIGNIFICANCE

Osteoarthritis is an age-related degenerative joint disease and affects a substantial part of the population over the age of 50. Total knee arthroplasty [TKA] is a common intervention, with a high rate of success. However, the optimal design of knee endoprosthesis is still under discussion, with a focus on transversal plane behavior, i.e. knee axial rotation [5,6].

METHODS

The FEF consists of a stiff lightweight aluminum arch with a compression screw (Fig. 1). Both sides of the arch are attached to individually molded thermoplastic shells over the femoral epicondyles. The FEF was tightened at a torque between 10 and 15 cNm. For the purpose of this study, the frame was mounted with a cluster set of three tantalum markers at the original location of the optoelectronic markers.

Seven patients (mean age 71 years), with a TKA and a BMI ranging from 26 to 30 were measured. The patients were asked to perform three step-up tasks, with the knee centered between the image intensifier and focus of a fluoroscope [4]. A three dimensional model of the TKA was fitted to the planar image to reconstruct the in vivo position and orientation of the TKA as well as the FEF markers [4]

The measurement error, defined as the difference between axial rotation of the prosthesis femoral component and the FEF marker set, was derived for each measurement time-frame. Accuracy and precision were calculated; a student t-test was used for comparison between modes.

RESULTS

At knee flexion angles above 35°, the knee sometimes moved out of the measurement volume. Results are lumped for each 10° of knee flexion and are shown in Table I. The RMS error appeared linearly dependent on the flexion angle. For knee flexion angles below 40° no systematic differences were found, while the precision is always better than 3.3°. The mean error over the whole range of knee flexion, i.e. 0°-40, is less than 2.2°

Knee Flexion Range	N	Mean error (Accuracy) Mean	SD	Absolute error (Precision) Min	Max	RMS
0°-10°	21	0.05	0.75	-1.21	1.31	0.89
10°-20°	21	0.04	1.83	-2.89	3.74	1.76
20°-30°	21	-0.08	2.62	-4.53	3.84	2.61
30°-40°	18	-0.67	3.58	-6.17	4.72	3.28
40°-50°	13	-2.17 *	3.71	-7.10	2.70	4.18
50°-60°	5	-2.86 *	4.92	-7.92	2.93	5.17

Table I Error of the measured axial rotation of the FEF, as a function of knee flexion angle. Nsubjects= 7; N = number of succesfull trials. All values in degrees, internal rotation is positive. An * indicates a significant difference. RMS= root mean square.

DISCUSSION

A linear dependency of RMS error with knee flexion angle was also found by Südhoff et al. [7], who used biplanar Xrays to evaluate a similar kind of device. However, their subjects reported pain and limping, while no side effects of wearing the FEF were reported by our subjects. Houck et al. [3] validated the use of a epicondylar clamp, against bone-pins in a study on walking of one healthy young man, and found an error of more than 5° for knee flexion angles below 40°. Based on a group of typical subjects with higher BMI's, we found errors of less than 3.3° for that range.

So it can be concluded that the FEF is performing superior over existing femural attachments. The resulting precision, i.e 2 till 3°, is in the same order of magnitude as differences that might be observed in comparing differrent knee endoprosthesis[5,6]. So, use of the FEF does not permit valid assessment of knee axial rotations at individual level. However, fluorscopy would not allow complex task evaluation, like turning away while rising from a chair, provoking knee axial rotation [8]. Therefore it is concluded that use of the FEF with optoelectronic stereometry is a valid method to track axial femoral rotation for knee flexion angles less than 40°, for most clinical studies.

ACKNOWLEDGMENT

Danny Koops is gratefully acknowledged for his assistance in engineering the FEF

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ESTABLISHING THE ACCURACY OF 3-D ULTRASOUND DETERMINATION OF THE HIP JOINT CENTRE WITH MRI

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Introduction

The validation of biomechanical models for gait analysis can be difficult. Kinematic fitting or functional calibration techniques can be validated by comparing predictions of hip joint centre location with imaging techniques. An advantage of kinematic fitting techniques is that they result in the prediction of the position of hip joint centre (HJC) in relation to bony landmarks of the pelvis. Previous work has compared kinematic results with radiographs [1], which are invasive, or MRI [2] which are expensive, but these are time consuming processes. Hicks and Richards [3] have suggested an apparent agreement between kinematic fitting and ultrasound measurements but they did not validate the ultrasound technique. This study extends the approach used by Hicks and Richards by using a full 3-D reconstruction of the ultrasound data and validating this technique against a gold standard of MRI scans.

Statement of Clinical Significance

This study provides a basis for the validation of new lower limb gait analysis kinematic modelling techniques. 3-D ultrasound could be used in evaluating the suitability of new kinematic models for clinical use.

Methods

Fourteen healthy subjects participated in the study with a mean age of 36 years (range 17 – 70 years) and BMI of 23 kg/m^2 (range 18 – 32 kg/m^2). Participants underwent a MRI scan with six series of PD-weighted Fast Spin Echo images in a 3T Siemens Trio (Erlangen, Germany). Participants then had an ultrasound scan of their pelvis to identify specific anatomical landmarks used for marker placement, and the femoral heads, using a Beamformer Echo Blaster 128-1Z (Vilnius, Lithuania). To determine the Hip Joint Centre (HJC) our protocol used the 2-D ultrasound images reconstructed in 3-D space. This determination is achieved by the simultaneous use of the ultrasound imager with a 3-D tracking system able to locate the position and orientation of the ultrasound probe. For this purpose, markers were rigidly attached to the ultra sound probe. Ultrasound measurements aimed to identify the location of the femoral heads in 3-D. From this data the 2-D ultrasound image can be reconstructed in 3-D.

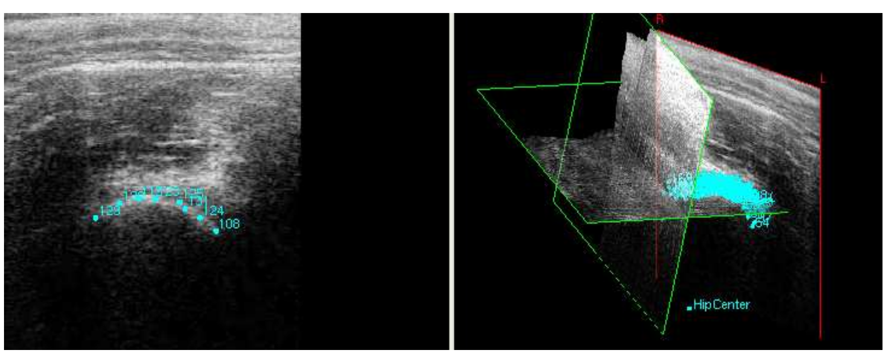

Figure 1: 3-D reconstruction of ultrasound and landmarks indicating the femoral head

The perimeter of the HJC was outlined by landmarks using Stradwin software (Figure 1) Sphere-fitting, using a Gauss-Newton method, was then used to determine the HJC from these points.

Results

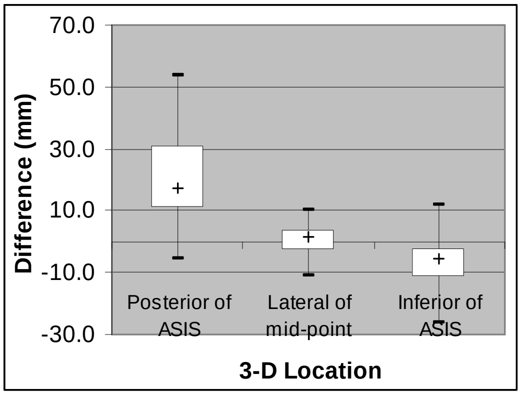

Figure 2: Difference in 3-D co-ordinates of MRI and ultrasound HJC

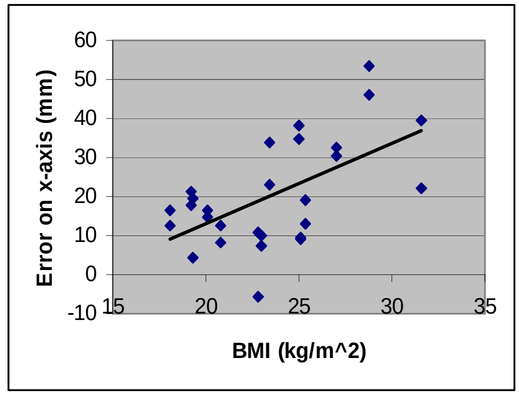

Figure 3: The relationship between x-axis error and participant BMI

Discussion

There was excellent agreement in the MRI and ultrasound determined distance between HJC where the average error was 3 ± 3.5mm. This result is promising as it is the primary measurement needed to validate lower limb gait models. It can also be seen that ultrasound has the potential to determine the 3-D location of the HJC close to the accuracy of MRI in the y- and z-directions. Indeed, the difference between the 3-D location determined by the gold standard MRI and the ultrasound method is 4±2mm in the y-direction and 9±6mm in the z-direction. In the y-direction this difference is minimal. Results in the x-direction, 21±13mm, indicate much greater error, however, there is a strong indication that this is dependant upon body mass index (Figure 3). It is thought that some of this error may also be attributed to the definition of the pelvic frame in the 3-D ultrasound. Additionally, it is acknowledged that the MRI scans are obtained in lying whilst the ultrasound scans are collected while the participant is standing. Further work is in progress to determine the precise reason for this and to investigate the potential for correcting this.

Conclusion

It is possible to use 3-D ultrasound to determine the position of the hip joint centre. The ultrasound determined distance between hip joint centers is in strong agreement with that obtained from the MRI scans. The 3-D location of the hip joint center is also good in participants with normal BMI. Further work is required to ensure that these 3-D ultrasound techniques are applicable to participants with higher than normal BMI.

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Comparison of techniques to determine the position of the hip joint centre against a CT-image gold standard: an *in vivo* study

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Introduction

The estimation of joint positions is an essential aspect in both clinical gait analyses and musculoskeletal modelling, where the location of the hip joint centre plays a key role in predicting the loading conditions of the lower limb. Although a variety of non-invasive approaches for the determination of centres of rotation have been proposed using [1], the comparative accuracy with which the hip joint centre can be determined in vivo remains unknown. The aim of this study was therefore to directly determine the position of the hip joint centre using both functional and regression techniques and to validate these predictions against the anatomical hip joint centre, as reconstructed using a CT gold standard.

Methods

Eight patients (BMI: 27.0 ±4.4) were recruited, 8-10 years after total hip replacement. After the attachment of reflective skin markers to the thigh and hip segments, CT scan imaging was performed to allow 3D reconstruction of the femoral head position relative to the skin markers. Finally, the patients were required to lie on a bed while the investigator moved their lower limbs passively during motion capture. The kinematic data sets and anthropometrical information were used for the functional estimation of the hip joint centres using functional methodologies (Symmetrical Centre of Rotation Estimation (SCoRE), Geometric Sphere Fit Method (GSFM)) [1] and regression methodologies [2] respectively.

Results

The SCoRE technique produced the most accurate estimation of the hip joint centre using functional methodologies (Figure 1). Seven centres of rotations were estimated within an accuracy of 11.9 and 18.7mm with the main error direction tending towards anterior and caudal. The GSFM estimated the centre within a range of 17.5 to 45.5mm. SCoRE was also able to predict the hip centre more accurately and with a smaller distribution than the Davis regression method (average error 22.9mm).

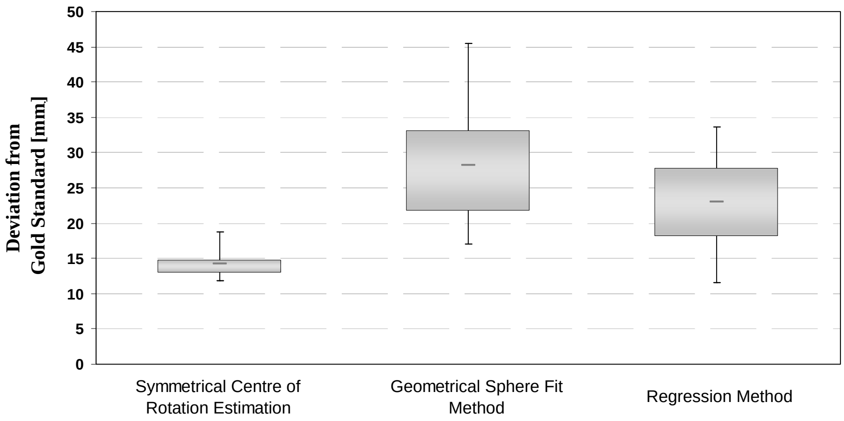

Figure 1: Absolute error in the estimated hip centres using both functional and regression methods, compared against the femoral head, as reconstructed from CT scan data, in 8 THA patients.

Discussion

For the first time, we have directly compared the SCoRE and GSFM methods against in vivo datasets, and compared these functional joint estimations against regression techniques, as proposed by Davis et al. [2], which are used as the current standard for approximating rotational joints in many commercial motion capture systems. Furthermore, the regression methodology examined in this study performed under optimal conditions, since direct anatomical landmarks were used, which were not subject to skin marker artefact. When such exact anatomical data is not available, however, further decreases in accuracy through mal location of bony landmarks can be expected [3].

Despite the occurrence of soft tissue artefacts, the SCoRE technique has been shown to allow a considerable improvement in accuracy over previous methodologies for the non-invasive determination of skeletal joint centres, using only motion data.

Acknowledgements

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Optimization of segment scaling and marker positions to drive a three-segment musculoskeletal foot model using gait motion analysis data

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Introduction:

Musculoskeletal models can be driven either by inter-segmental joint angles or directly by measured marker trajectories. Driving the model by marker trajectories involves the assumption that markers are placed at accurate landmarks as specified in the model definition, and in turn ignores the inherent error in marker motion during the capture session. Driving the model by pre-calculated inter-segmental joint angles neglects any translation at the joints. A routine was developed to optimize the bone segment scaling and marker location to satisfy the assumed spherical and revolute joint constraints for a three-segment foot model. The musculoskeletal foot model used had the same segmentation of the foot used in the kinematic model.

Clinical Significance:

A musculoskeletal model driven by gait motion analysis data can be developed into a useful clinical tool to assess gait pathologies. Calculated inter-segmental angles of the foot and ankle are highly sensitive to marker error due to the close proximity of markers. An optimization method which reduces skin motion errors may enhance reliability and improve accuracy of these models, leading to increased clinical utility. Such models can also be used to model the muscle tendon transfer surgery to predict the outcome of intervention.

Methods:

A clinical foot model for gait analysis utilizing retroreflective markers and a motion capture system was applied to three normal children. A 16 degree of freedom kinematic leg model comprised of the thigh, shank, hindfoot, forefoot and hallux was developed to match the segments defined in the motion model. The knee, ankle and forefoot/hindfoot joints were modeled as spherical joints and the hallux/forefoot joint as a revolute joint. An optimization routine was constructed to scale each segment length as well as the marker positions to minimize the total sum of squared differences between the segment fixed markers on the model and the measured marker trajectories over the whole trial[1,2].

Once the segment scaling and marker position on the bone segments were optimized to match the motion capture data, the optimized marker position in the global reference frame were saved. The sum of squared differences between the trajectories of the optimized segment-fixed markers and originally captured markers is called total marker error. Inter-segmental angles were calculated using both (optimized and original capture) marker data and the difference of these two measures is scaled and plotted with the total marker error in figure 1(b) to show the correlation between the two.

Results:

Figure 1(a) shows the inter-segmental ankle flexion angle during a gait cycle by using the raw marker trajectory data and same angle calculated by using the optimized marker trajectories. Even though the optimization routine makes the graph smoother, the average difference in inter-segmental angle was reasonably small (2^{o}-3^{o}). Average difference in graph peaks over all trials was also small (2.3^{o}) showing that the optimization did not affect the gross motion of joints. The difference in the two plots can be explained by comparing the total marker error, in the inter-segmental angles over a gait cycle, with the total errors in all the marker trajectories over a gait cycle as shown in figure 1(b). Table 1 shows the correlations between the total marker error and three inter-segmental angles: hindfoot relative to shank, forefoot relative to hindfoot and hallux relative to midfoot. The same process was conducted for the walking trials of three normal pediatric subjects.

All the trials showed a significant but moderate correlation between the total marker error and angle errors, which explains that most changes in the inter-segmental angles are caused by the change in marker trajectories. These changes in marker trajectories are required to satisfy the enforced joint constraints as well as the assumption that the markers are rigidly attached to the segments.

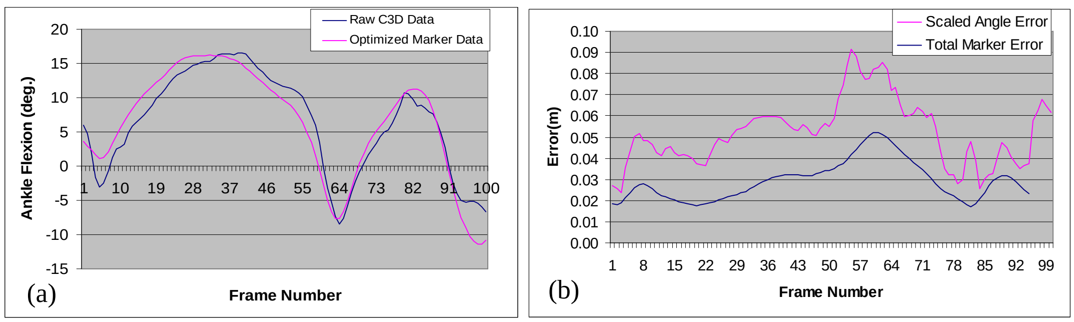

Figure 1: (a) Comparison of ankle angle. (b) Comparison of marker and angle error.

Inter-Segmental Angle	Correlation with total marker error					
	Subject 1		Subject 2		Subject 3	
	r	p	R	p	r	p
HindFoot/Tibia	0.332	0.001	0.613	<0.001	0.261	0.007
MidFoot/HindFoot	0.305	0.002	0.430	<0.001	0.252	0.010
Hallux/MidFoot	0.475	<0.001	0.404	<0.001	0.213	0.030

Table1: Correlation between total marker error and inter segmental angle.

Discussion:

The results of this study show the efficacy of an optimization routine to enforce model joint constraints by adjusting the segment scaling and local marker coordinates. The process does not affect the gross motion of the segments significantly as observed by the relatively small changes in joint angles peaks. The differences between the two analysis results can be attributed to soft tissue artifacts, the introduced simplified joint constraints and the assumption that the markers are rigidly attached to the segments. The optimized model results in smoother joint angle curves and, hence, unrealistic accelerations are reduced. This is important for a musculoskeletal model since the equilibrium equations depend directly on the accelerations.

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Gait of Children with/without Cerebral Palsy; Work, Energy, and Angular Momentum

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INTRODUCTION

Previous efforts to quantify the work done in walking has concentrated only on the external work[1,2], or work done to move the body center of mass. External work only explains approximately 50% of the work done in walking Willems et al[3]. Recent work has suggested that angular momentum, like energy is "conserved to a large extent" and could be a control parameter in walking. Yet, there has been little data presented to validate this claim and no one has explored angular momenta or the implications with respect to biomechanics, motor control, and energy of gait in children or children with CP.

In this study we examined the work, (internal/external), energy, and angular momenta of TD children and children with CP at preferred walking speeds. Our hypotheses were: 1) the internal work represented a significant portion of the overall work of both normal and pathologic gait, 2) angular momenta would be greater in children with CP compared to TD children, and 3) the angular momentum of the body can be used as an indicator to quantify the contribution of individual limbs and segments to the total internal work done over a gait cycle.

CLINICAL SIGNIFICANCE

Children with CP typically have shorter strides, slower preferred walking speed, reduced range of motion of the joints, and altered ankle moment patterns as a result of the lack of a true heel strike. In addition their gait requires two to three times more mechanical work to travel the same distance, is less pendular with reduced energy recovery, and is thus more metabolically expensive. This study attempts to quantify the gait motions associated with this increased cost of walking in children with CP.

~~Subject specific human walking simulation provides greater understanding of walking patterns for different situations (i.e change of walking speed) without requiring additional experimental data. Applications include gait analysis, gait optimization, and determination of properties for assistive devices, i.e. braces, walkers, etc.~~

METHODS

Kinematic data on a convenience sample of 24 children were collected and analyzed. This group of children consisted of two populations. The first group of age-matched controls was comprised of 8 children without known musculoskeletal, neurological, cardiac, or pulmonary pathology. The second group consisted of 16 children diagnosed with spastic diplegic CP, these subjects were community ambulators and walked without aids.

For each subject, the angular momentum about the body CoM was calculated as the sum of each segments total momentum which is itself the sum of the local segment moment of inertia and the segments moment of momentum about the body CoM. Next the whole body work/energy was calculated, again this is the sum of the work/energy, both internal (work about CoM) and external (work on Body CoM) of each segment allowing energy transfer between limb segments only.

To determine the relative energy due to motion in each direction, a regression was conducted using the absolute value of the momentum about the other two axis, i.e. energy in X direction related to momentum about the Y and Z axis. The absolute value of the momentum was used to illustrate the positive energy associated with both positive and negative momenta. The relative energy represents linear component of rotational motion, and momenta direction is represented normal to the plane of rotation so directional energy is a result of the rotational motion in the two orthogonal planes, hence the regression with out-of-plane momenta.

RESULTS

When walking at their CWS children with CP did 59% more total work per unit mass and distance traveled than the TD group, 1.64 J/kg-m and 1.03 J/kg-m respectively, P>0.01. For CP subjects 60% of this total work is attributed to external work or movement of the body CoM, while external work accounted for 47% of total work for TD, or 0.99J/kg-m and 0.49 J/kg-m respectively, P<0.01. While walking, children with CP did 21% more internal work than TD children, P=0.053. For children with CP this results in 0.66 J/kg-m of internal work being done vs. 0.54 J/kg-m for TD children, 40% and 53% of the total work respectively.

About the Z axis, the absolute area under the curve, momemtum area, was 73% larger in the group with CP, P<0.01. The momentum area about the X axis reflect the side to side leaning motions of the body and is larger in the group with CP by 50%, p=0.058.

In total body analysis the relative energy in the X and Z directions correlated with the associated momentum for both groups, TD: R^2=0.75 and 0.95 respectively, P<0.0001, and CP: R^2=0.88 and 0.72 respectively, P<0.0001. There was a poor correlation of the relative energy in the Y direction, TD: R^2=0.33 P<0.0001 and CP: R^2=0.09 P<0.0035.

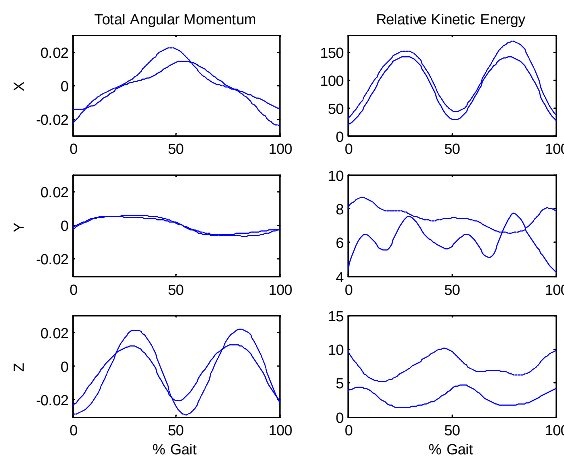

Figure. Typical whole body Angular Momentum and relative kinetic energy. Solid-normal, Dashed-CP

CONCLUSIONS

In analysis of gait and the metabolic cost of locomotion, internal work offers a rich area for analysis for increasing efficiency of walking. This is particularly relevant when looking at pathological gait such as that associated with CP since it represents motion with no direct application to forward motion. In addition, the angular momenta about the body CoM offers insight into the dynamic motions associated with internal work. However, we see hat the angular momenta are well organized within both groups, suggesting that changes in the momenta of an individual segment will result in changes in other momenta. While global analysis offers indirect insight into gait, a more detailed analysis of body limbs or segments offers a better cause/effect relation between gait kinematics and work.

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Use of Gait Analysis and Physical Exam Data for Surgical Decision Making in the Correction of Equinus Gait Deformity in Children with Cerebral Palsy

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Introduction

Equinus gait is a common condition seen in ambulatory children with spastic cerebral palsy. Tendo-Achilles lengthening (TAL) and intramuscular gastrocnemius lengthening are two surgical methods commonly employed to correct equinus gait. Both physical examination and instrumented gait analysis (IGA) data are routinely used to make decisions regarding which surgical procedure will best benefit a patient, but no concrete information exists for selecting one procedure over another. The Silfverskiöld test, often included in the physical exam, has been suggested as a valuable measure to establish soleus involvement in the contracture and assist in surgical decision making.[1,2] The purpose of this retrospective study was to evaluate the selection criteria for TAL and intramuscular lengthening by comparing surgical outcomes using pre/post changes in gait and physical exam measures from IGA.

Clinical Significance

There is concern that the intramuscular lengthening does not provide sufficient correction resulting in recurrence, while the TAL provides excessive correction contributing to iatrogenic crouch gait. Choosing the appropriate surgery for a particular patient can be difficult, so quantitative gait performance data may be used as clinical evidence for the decision. This study uses instrumented gait analysis data to evaluate selection criteria and surgical outcomes for each procedure. Results may provide objective criteria for determining the most suitable intervention, decreasing surgical risk while enhancing clinical benefits.

Methods

This study retrospectively reviewed gait analysis data of children with spastic CP aged 3-14 with equinus gait who underwent a percutaneous TAL performed using three partial tenotomies, or an intramuscular lengthening (STR) performed as described by LM Strayer.[3] All subjects had a pre- and post-operative gait analysis between August 1999 and August 2007. Exclusion criteria for this study were prior calf lengthening procedures, dorsal rhizotomy or baclofen pump placement, simultaneous midfoot osteotomies, and neurotoxin injections at the calf between pre- and post-operative gait analyses. All subject data were collected at our institution under a protocol approved by the Institutional Review Board. Comparisons were made between the two surgical groups for 6 kinematic variables: average dorsiflexion throughout the gait cycle (DFGC), average dorsiflexion during stance (DFSt) and swing (DFSw), dorsiflexion at initial contact (DFIC), maximum dorsiflexion during stance (DFMax), the timing of DFMax (PGCDFMax). In addition, two physical exam variables were evaluated, passive dorsiflexion with the knee flexed (PE:DFKF) and with the knee extended (PE:DFKE). Differences between the groups were analyzed using Student's t-test.

Results

This study analyzed data from 42 subjects (67 limbs), of which there were 16 hemiplegic, 22 diplegic, 3 triplegic and 1 quadriplegic subjects. Thirty-four subjects were GMFCS I, 4 were GMFCS II and 4 were GMFCS III. Average follow-up time was 18 months (±12 months). There were statistically significant differences found for each variable tested pre- to post-operatively for both procedures with the exception of PE:DFKF and PGCDFMax in the STR group (see table 1). Between group comparisons were significant for a difference in the pre-operative DFSt (p=0.039), pre-operative PE:DFKF (p=0.006) and post-operative PGCDFMax (p=0.014). The amount of pre- to post-operative change between the two groups showed a significantly greater difference in the TAL group for DFGC (p=0.030), DFSw (p=0.035), DFMax (p=0.033), PGCDFMax (p=0.016) and PE:DFKF (p=0.003).

	TAL (n =43 limbs, n=29 PE:DFKF only)			**STR** (n=24 limbs, n=22 PE:DFKF only)			**TAL vs. STR**		
Variables	**Pre**	**Post**	**Δ**	**Pre**	**Post**	**Δ**	**Pre – Pre Δ**	**Post – Post Δ**	**TAL Δ – STR Δ**
PE:DFKE (°)	-4.7 ± 10.0	10.2 ± 12.0	14.9*	-1.5 ± 8.4	8.5 ± 10.4	10.1*	3.1±9.4	1.7±11.5	4.8±13.0
PE:DFKF (°)	-0.7 ± 11.5	12.0 ± 9.4	14.4*	6.0 ± 7.2	9.1 ± 8.4	3.9	6.7±10.1*	2.9±9.0	10.5±11.7*
DFGC (°)	-8.4 ± 14.3	5.4 ± 7.6	13.8*	-3.5 ± 6.7	3.9 ± 7.4	7.4*	4.9±12.2	1.5±7.5	6.4±12.7*
DFSt (°)	-4.4 ± 14.3	8.9 ± 6.3	13.3*	1.3 ± 7.8	8.5 ± 7.0	7.2*	5.7±12.4*	0.5±6.6	6.2±12.5
DFSw (°)	-14.9 ± 15.1	0.1 ± 10.3	15.0*	-10.4 ± 8.2	-2.3 ± 9.5	8.1*	4.5±13.0	2.4±10.0	6.9±14.5*
DFIC (°)	-8.8 ± 11.7	0.2 ± 8.9	9.0*	-4.5 ± 7.7	3.1 ± 8.8	7.7*	4.3±10.5	3.0±8.9	1.3±11.4
DFMax (°)	4.4 ± 13.1	16.7 ± 7.1	12.3*	10.0 ± 9.7	15.3 ± 8.6	5.4*	5.6±12.0	1.3±7.7	6.9±12.5*
PGCDFMax (%)	23.5 ± 14.7	37.0 ± 14.4	13.4*	24.2 ± 14.0	27.7 ± 14.3	3.5	4.2±10.5	9.2±14.4*	10.0±15.8*

Table 1 – The mean ± standard deviation for the kinematic and physical exam variables pre-operatively, post-operatively and the change following surgery. *-p<0.05 for comparison pre- to post-operative values within procedure groups.

Discussion

Our results indicate that the Silfverskiöld physical exam test and DFSt are used for surgical decision making in the treatment of equinus gait at our institution. Patients with more soleus involvement and greater stretch response of the gastrocnemius-soleus complex were selected for a TAL. No other differences were found suggesting that those two variables are most important in our decision making process and our selection criteria were appropriate. Similar post-operative outcomes between the groups indicate that TAL resulted in better reduction of the contracture in the gastroc-soleus in those patients with more severe pre-operative equinus. Improvement towards normal in PGCDFMax following TAL suggests that there is a greater effect on reducing the stance phase stretch response. No difference in the change of DFSt and the fact that DFMax and PGCDFMax approached but did not exceed normal reduces concern for short term risk of over-lengthening. Results from this study allow us to begin constructing selection criteria for TAL and STR. Based on this data, we feel confident that those patients with soleus involvement indicated by the Silfverskiöld test and greater calf stretch response during gait will show greater improvement in gait following TAL, specifically with respect to swing phase dorsiflexion and timing of DFMax. Further study of ankle kinetics and knee and hip kinematics will help us further refine the selection criteria for TAL versus STR.

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Gait Analysis for Treatment Decision-making in Persons with Multiple Sclerosis: A Case Study Using Kinematics, Kinetics and Electromyography

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PATIENT HISTORY

The patient was a 38 year old woman with multiple sclerosis (MS) diagnosed in 1995. She has had no previous lower extremity surgery and is not wearing an ankle-foot orthosis (AFO). Her major issues include lateral left ankle instability and clearance problems during gait. She was referred for gait analysis to determine specifics for possible Botox injections to minimize spasticity that had been noted in the quadriceps, plantar flexors and posterior tibialis on clinical assessment. It was proposed that spasticity in these muscle groups was limiting left side range of motion and clearance.

CLINICAL DATA

Asymmetry in muscle strength was noted with typical strength on the right side and global weakness on the left. Left ankle dorsiflexion (1/5), plantar flexion (2/5), forefoot eversion (2/5) and inversion (4/5) were all less than typical values. Knee and hip flexion were 3/5, knee extension 5/5 and hip extension 4/5 (knee flexed and extended). Mild tightness was noted in the left ankle plantar flexors (10° with knee at 90° and 5° with knee at 0°). Minimal spasticity was noted in the left hamstrings, plantar flexors, (including sustained clonus) and posterior tibialis. A positive Ely was noted in the left rectus femoris (2/4). Spasticity for the above muscles was confirmed with simultaneous electromyography (EMG) recordings.

GAIT DATA

Visual evaluation of the patient's gait showed an unstable gait pattern with reduced walking velocity (range: 53 to 85 m/s) and sagittal plane range of motion.

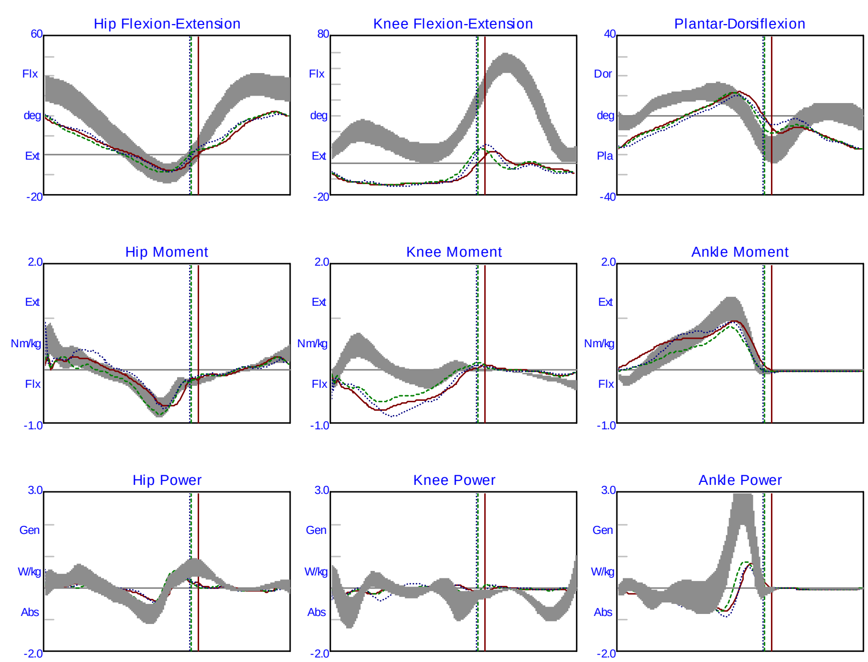

Figure 1: Sagittal hip, knee and ankle kinematics and kinetics during barefoot walking for three strides of the left side. (Typical reference = gray band)

Sagittal plane left hip, knee and ankle joint kinematics and kinetics during barefoot walking (Fig. 1) showed issues at all levels:

1) Reduced hip range of motion and reduced hip power generation at toe off
2) Knee hyperextension in stance with associated knee flexor moment
3) Minimal knee flexion in swing with overall minimal knee range of motion
4) Excessive ankle equinus in terminal swing/initial contact with associated plantar flexor moment in loading response
5) Delayed ankle dorsiflexion
6) Reduced ankle power generation in terminal stance

Electromyographic data did not support spasticity in the muscles tested **during ambulation**. Minimal activity was noted in the gastrocnemius and anterior tibialis during their respect phases and no activity was noted in the rectus femoris in swing (Fig. 2).

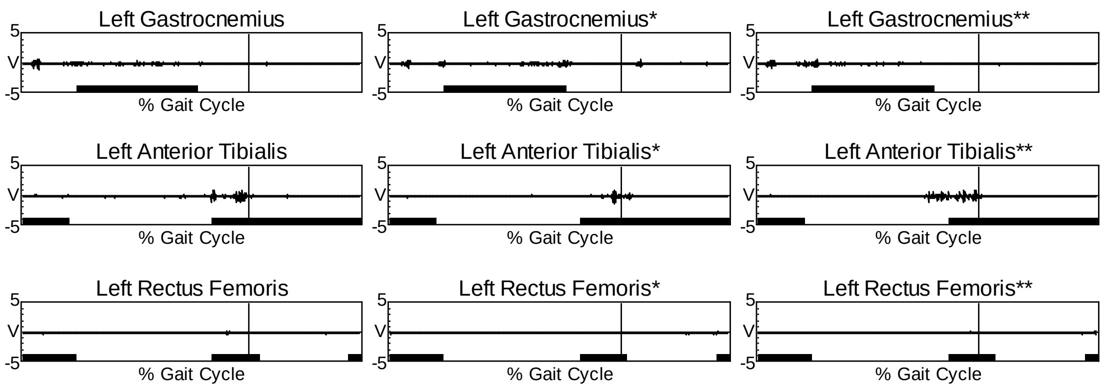

Fig. 2: Raw EMG signal for 3 gait cycles for the left gastrocnemius, anterior tibialis and rectus femoris. Expected muscle activity is provided by the black bars.

TREATMENT DECISIONS AND INDICATIONS

1) No Botox injections at this time: Electromyographic data suggests that Botox injections to the proposed muscles would have minimal effect on gait as these muscles are showing minimal contractions during gait as noted on dynamic EMG.
2) A custom AFO set in minimal dorsiflexion: to correct for the drop foot in swing as noted on ankle kinematics which will also help to reduce knee hyperextension and associated knee flexor moment also noted on knee kinematics and kinetics.
3) The patient has sufficient knee extensor strength to support a small knee extensor moment which may occur as a result of an AFO molded in slight dorsiflexion.

SUMMARY

Kinematic and kinetic data at the knee during barefoot walking showed knee hyperextension and an associated knee flexor moment in stance as a result of an excessive plantar flexion knee extension couple. As a result there was delayed dorsiflexion in stance. Peak knee flexion in swing was substantially less than typical due to reduced ankle power generation in terminal stance and hip power generation at toe off related to muscle weakness. Electromyographic data showed no activity of the rectus femoris in swing phase indicating that this muscle did not contribute to reduced knee flexion. The drop foot in swing was related to an absence of anterior tibialis contraction in swing and not activity of the gastrocnemius in swing. The kinematic, kinetic and EMG data indicated that the primary cause of decreased clearance in swing during ambulation was muscle weakness, specifically of the ankle plantar flexors and hip flexors, and not spasticity.

How do MRI brain lesions relate to gait characteristics and motor deficits in children with Cerebral Palsy?

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INTRODUCTION
A wide range of brain lesions leads to cerebral palsy (CP), one of the most common motor disorders in children. So far only general relationships between brain lesions and motor outcome in children with CP have been described. This pilot study examines the gait pattern and motor deficits in detail and relates them to four categories of brain lesions in CP.

CLINICAL SIGNIFICANCE
A more thorough study of the relationship between brain lesions and motor outcome in children with CP would enable individually tailored treatment strategies and might even enable the prevention of certain adverse outcomes.

METHODS
In 43 hemiplegic patients, between the age of 4 and 15 (mean: 9y, SD: 3y), gross motor function was evaluated using GMFCS. Gait patterns were classified according to Winters & Gage [1] and 31 clinically relevant 3D-gait parameters (kinematics, kinetics and video data) were defined. Spasticity, strength and selectivity were evaluated through ‘overall scores’. These overall scores were averaged subscores of the clinical exam per muscle group. In addition, an extra overall score combining muscle strength and selectivity was calculated. Furthermore, the number of involved spastic joint levels, weak joint levels and joint levels with limited selectivity was registered. Finally, ‘overall gait deviation’ scores reflecting general pathology of gait were registered. These were root mean square deviations from averaged age related typical gait trials. Brain MRI-data were available for all patients, with age at time of the scan ranging from 5 days to 15 years (mean: 5y, SD: 4y). MRI-data were classified according to Cioni into the following categories of brain lesions [2]: 1) malformations, 2) periventricular white matter lesions (PW), 3) cortico-subcortical lesions (C-SC) and 4) encephaloclastic lesions (EC).

RESULTS
Compared with the C-SCgroup, muscle strength was less impaired in the PW group ($p=0.019$). Children with EC lesions on the other hand, were more proximally involved as the pelvis tilted more anteriorly when compared with children with malformations, PW or C-SClesions ($p=0.003$, 0.005 and 0.020 resp.). *(Table 1)*

Table 1: Mean and standard error of parameters that were significantly related to brain lesions.

Brain lesion	Mean Pelvic tilt (°)	Muscle strength score (1 – 5)
1	12.62 (± 3.92)	3.83 (± 0.11)
2	17.81 (± 0.88)	3.62 (± 0.07)
3	18.45 (± 1.26)	3.29 (± 0.12)
4	23.98 (± 1.16)	3.28 (± 0.13)

Brain lesion 1: malformations, 2: periventricular white matter lesions, 3: cortico-subcortical lesions, 4: encephaloclastic lesions

In the malformations group, children more frequently (χ^2) exhibited a normal pattern instead of the typical double-bump pattern often observed in the ankle-moment graph ($p=0.017$).

No relationship was found between brain lesions and the ‘overall gait deviation’ score. However, a significant relationship was found between gait pattern and the ‘overall gait deviation’ score. Children classified as Winters & Gage H2 were generally more involved than children classified as H1 and A1 (p= 0.005 and 0.025 respectively) as reflected by their ‘overall gait deviation’ score while children classified as K2 were generally more involved than children classified as A1 (p= 0.036). Furthermore, children with a higher GMFCS ranking tended to be more involved as reflected by their ‘overall gait deviation’ score (p=0.052). *(Table 2)*

Table 2: Average and standard error of ‘overall gait deviation’ scores for Winters & Gage’s gait classification and GMFCS classification

	Winters & Gage						GMFCS	
OGD	A1	A2	K1	K2	H1	H2	I	II
Average	581.79[1, 2]	745.81	/	807.56[2]	687.86[1]	865.23[1]	707.57	836.17
SE	53.00	37.59	/	91.77	41.97	83.66	28.52	63.53

OGD: overall gait deviation score, SE: standard error,[1, 2]: comparisons with p<0.05

DISCUSSION

Pelvic tilt, pattern of ankle moment and muscle strength were found to be significantly related to brain lesions. These findings could be a first step for early prediction of motor outcome in children with CP based on MRI brain images thereby facilitating individually tailored treatment strategies. So far only limited prediction is possible and further research based on detailed brain images and gait assessment is required in order to generate detailed and clinically useful predictions. Although we could not recognize a significant relationship with brain lesions, the ‘overall gait deviation’ scores were able to distinguish between gait patterns according to Winters & Gage and tended to distinguish between overall functional levels (GMFCS). The ‘overall gait deviation’ score thus proved to be a good measure of gait pathology reducing the redundant 3D-gait data to one single practical measure.

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AKNOWLEDGEMENTS

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CLASSIFICATION OF GAIT PATTERNS IN CHRONIC POST STROKE PATIENTS.

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Summary

The results of gait analysis data showed that gait after stroke deviates strongly from typical gait and presents a variety of patterns. In this study of 42 patients, three different patterns could be observed.

Conclusions

Gait following stroke differs significantly from typical gait. Three different patterns could be observed. The knee extension thrust pattern (N=19) and flexed pattern (N=16) resulted in abnormal kinematics and kinetics at all levels and a lower walking speed, while for the mild group (N=7) gait pathology was limited to significant deviated ankle kinematics and kinetics, with almost normal knee motion.

Introduction

Many authors described gait abnormalities, as well as classifications of gait patterns after acute and chronic stroke. Gait after stroke differs from typical gait at all levels and there is marked intersubject variability. Characteristic gait patterns after stroke often distinguish on knee kinematics (flexion – extension) and walking speed (fast – slow). The aim of this study was (1) to define the different gait patterns (known from literature) in chronic stroke, based on qualitative manual template matching of gait analysis plots and (2) to document objectively the characteristic gait parameters in those patterns.

Patients/Materials and Methods

42 patients with chronic cerebrovascular accident (CVA) (23L, 19R) with a mean age of 55 years (+/- 13.3) were included in this study, as well as 10 age-related control subjects. Mean time since CVA was 5.2 years (+/- 3.9y). Lesions were classified as an infarct in 23 patients and a haemorrhage in 11 patients. In 8 patients, details of the lesion were not known. All subjects underwent a lower limb 3D gait analysis, with or without walking aids, including kinematic and kinetic data (8 camera VICON system and 2 AMTI forceplates), EMG of 8 lower extremity muscle groups and a full clinical evaluation (ROM, spasticity, strength and selectivity). Three representative trials were selected bilaterally. For each subject, 58 clinically relevant gait parameters were defined and mean data and SD of these parameters were calculated. ANOVA analysis (post hoc Kruskal Wallis) was used for comparison between pathological and normal reference data and to examine the overall differences in parameters between observed gait patterns.

Results

Gait following stroke significantly deviated form normal walking at all levels; including reduced walking speed and step length, decreased hip ROM in the sagittal plane, decreased

knee flexion in swing and decreased power generation at hip, knee and ankle ($p<0.0005$). Based on literature and visual inspection, the knee was recognized as the most discriminating factor between patterns. Three different gait patterns could then be observed: (1) knee extension thrust pattern (N=19), (2) flexed pattern (N=16) and (3) mild pattern (N=7). The latter had normal knee flexion during loading response and showed no significant differences in hip and knee kinematics and kinetics compared to normal. Group 1 was characterized by a knee extension thrust pattern leading to knee hyperextension (in 17/19 patients) and an early knee flexion moment in stance. Knee flexion during loading response was absent ($p<0.0001$). The second group showed increased knee flexion at initial contact, decreased maximum knee-extension in stance and decreased knee flexion moment ($p<0.005$). Significant decreased hip ROM in the sagittal plane and decreased peak knee flexion in swing was seen in group 1 and 2 ($p<0.0001$). At the ankle, significant lower ROM at push-off was seen in all three groups. In the first and second pattern, a significant decreased walking speed ($p<0.0001$) was observed, whereas walking speed of the mild group was close to normal. Circumduction, as a compensation for reduced foot clearance, was most often seen in group 1 (18/19) and group 2 (10/16). Pelvic hiking was observed for 17 out of 19 patients in group 1 and 11 out of 16 patients in group 2. In group 3, 3 out of 7 patients showed circumduction and pelvic hiking.
Table 1 summarizes crucial significant parameters.

	Control data	**Total group**	Knee ext group	Knee fl group	Mild group
Walking speed (m/s)	1.32 (0.084)	0.46 (0.31)	0.41 (0.29)	0.38 (0.27)	0.77 (0.24)
Step length (m)	0.70 (0.048)	0.39 (0.15)	0.40 (0.16)	0.33 (0.12)	0.49 (0.10)
Hip ROM sagittal plane (°)	45.05 (3.53)	24.27 (12.0)	23.78 (12.65)	20.12 (10.13)	35.10 (7.21)
Max knee flexion swing (°)	58.74 (3.15)	33.57 (14.26)	31.09 (14.06)	31.08 (14.63)	46.0 (6.78)
Max knee ext stance (°)	1.77 (5.15)	0.35 (11.58)	-8.61 (7.43)	11.74 (6.92)	-1.36 (5.28)
Power generation ankle pushoff (Nm)	4.03 (0.65)	0.75 (0.99)	0.38 (0.47)	1.00 (1.40)	1.20 (0.50)

Table 1: mean data and standard deviations of 6 parameters.

Discussion

The majority of the differences between chronic post stroke gait and typical gait found in this study were consistent with those found in literature [1]. The three different patterns could be compared to some of the patterns described by Mulroy et al [2] and Kinsella et al [3]. Both research groups described a flexed and an extended gait pattern with slow walking speed. The mild group could be compared to their 'fast walking' group, with a more normal gait compared to the other subgroups. The study of Kim et al [1] also described the knee flexor moment pattern, comparable as the knee extension pattern found in this study.

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3-Dimentional Gait Analysis of Aquatic Treadmill Walking Compared to Conventional Pool Walking in People Post-Stroke

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INTRODUCTION

Overground and treadmill walking are two most common gait training modes for post-stroke rehabilitation. The use of treadmill for post-stroke has been documented to be effective in improving various gait parameters. Unique properties of water provide various benefits for gait training, such as buoyancy for weight support. Recent studies investigated the differences between overground and aquatic treadmill walking. However, few studied the differences between aquatic treadmill walking (ATW) and conventional pool walking (CPW). The purpose of this study was to compare biomechanical differences between ATW and CPW in people with and without stroke.

CLINICAL SIGNIFICANCE

The study findings will provide scientific understanding of ATW and CPW on gait patterns of people with stroke. The results of this study will give evidence-based guidelines for aquatic gait rehabilitation.

METHODS

22 adults with stroke (66.0 ± 11.1 years) and 22 healthy adults (60.1 ± 10.1 years) were tested for their walking patterns at a matched walking speed on two conditions: a) walking on a aquatic treadmill (Aqua Gaiter, FERNO, Ohio, 2002) and b) walking on a pool floor. A movable floor pool (KBE Bauelemate, GmbH & Co., Germany, 2002) was used to provide consistent pool depth adjusted to participant's xyphoid process at the chest. All participants had to complete three testing trials for each walking condition with 1-minute rest time between trials. All testing trials were captured by six underwater lenses (Underwater Camera Company of America, San Diego, CA, 2005) connected to six digital video cameras (Canon, Japan, 2006) outside of the pool. All trials were synchronized with an audio sound. Vicon Motus Video v.9.2 (Vicon, Oxford, UK, 2008), was used to digitize, process, and analyze gait variables (spatiotemporal, peak joint angles and joint excursions of hip, knee, ankle). 15 water-proof reflective markers were placed on the bony landmarks of the lower extremities.

RESULTS

Comparisons using paired t-tests revealed that each group had significant differences between the two walking modes in spatiotemporal and joint kinematic variables. The stroke group demonstrated differences in 3 gait variables in the involved side and in 4 variables in the less-involved side (Table 1). Whereas, the non-stroke group showed changes in 2 variables on each side (Table 1). Repeated measures mixed model ANOVAs showed significant group interactions in 7 dependent variables ($p<0.05$): stride length, foot off percentage, peak hip extension, peak hip abduction, peak hip adduction, hip flexion/extension excursion and knee flexion/extension excursion. After Bonferroni correction ($p<0.05/20$) only 5 gait variables showed significant group interactions.

		Involved/Right Side		*p*	Less Involved/Left Side		*p*
	Walking Mode	**ATW**	**CPW**		**ATW**	**CPW**	
	Variables	**Mean ± SD**	**Mean ± SD**		**Mean ± SD**	**Mean ± SD**	
SG	**Stride Length (m)**	0.64±0.19	0.68±0.24	0.156	0.63±0.18	0.72±0.28	0.007*
	Hip Abd (º)	5.50±8.08	9.21±7.85	0.031*	7.70±6.84	11.90±6.82	0.001**
	ROM Hip Abd/Add (º)	10.05±4.68	13.08±4.42	0.007*	10.11±3.52	12.76±4.42	0.004*
	ROM Hip Int/Ext Rot (º)	27.82±9.88	36.16±10.86	0.001**	25.82±7.72	35.94±12.84	0.001**
NSG	**Stride Length (m)**	0.81±0.18	0.96±0.13	0.000**	0.81±0.19	0.99±0.12	0.000**
	ROM Hip Flex/Ext (º)	42.27±8.05	47.62±8.90	0.014*	42.69±7.08	49.06±7.98	0.003*

Table 1: Gait parameter comparison summary.
***Statistically significant (P<0.05),**
****Statistically significant with Bonferroni correction (P<0.05/20)**

DISCUSSION

People post-stroke demonstrated decreases in peak hip abduction, hip abduction/adduction excursion and hip rotation excursion of both legs during aquatic treadmill walking. This may help them reduce the leg circumduction during aquatic walking. It seems that less water resistance during aquatic treadmill walking contributes to having such decreases as compared to conventional pool walking. Meanwhile, healthy adults showed decreases in stride length and hip flexion/extension excursion of both legs. It appears that stationary walking position and more upright trunk position during aquatic treadmill walking help them not to exaggerate their stride length and hip motion. As for the group interactions, the results were somewhat mixed. Stroke group showed greater reductions associated with aquatic treadmill walking in peak hip extension and peak hip abduction. During aquatic treadmill walking healthy adults revealed greater rate of decreases in stride length, peak hip adduction, and knee flexion/extension excursion. Hip flexion/extension excursion showed an opposite direction of changes across the two walking modes: people post-stroke increased it while healthy adults decreased it. A possible explanation for this mixed group interactions is that people post-stroke were required to compensate for different walking modes while health adults were able to actively adjust their walking to them. The differences in their motor adaptability may be related to the mixed results. In conclusion, aquatic treadmill can provide more ideal gait training environment for people post-stroke or similar neuromuscular conditions as compared to conventional pool walking.

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The comparison of changes in the body shape in American and European children, with the application of age invariant human body shape index HBSI.

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Introduction

Human Body Shape (HBS) is usually monitored with the age and sex specific Body Mass Index (BMI).The Human Body Shape Index (HBSI) has been introduced recently as a sex specific measure of HBS. The HBSI offers an alternative to the standard age grouping of BMIs, and allows for characterization of HBS for an entire group using one index which is independent of age. [1] The HBSI provides faster, cheaper and more sensitive method, than BMI to monitor the changes of HBS in individuals and to compare the changes in HBS (including obesity) in different pediatric populations. The objective of this study was to estimate the changes in HBS of American and Polish children (5-19 years old) using HBSI, over 20 years (1985-2006).

Statement of Clinical Significance

HBS is the primary determinants of body structure. The simplest interpretation of the second Newton's law reveals the direct effect of the object's mass on its kinematics. The obesity related changes of HBS affect functional performance in adults in children, constituting an important confounding factor in the gait analysis, which should be addressed with appropriate normalization techniques. The application of HBSI instead of BMI to account for HBS changes in children who undergo gait evaluation should decrease the variability of gait parameters, thus increasing theirs sensitivity to detect impact of any intervention.

Methods

Data consisting of body mass M(kg) body height H(m), sex and age (years) of US and Polish children were obtained for omparable time periods (table 1). The data for US children were obtained from the National Health and Nutrition Examination Survey (US_1,US_2,US_3) (NHANES).[2] In Poland, data were collected from school-age children in Warsaw, Poland area (PL_1 , PL_2). The HBSI were calculated as M/H^{χ} for girls ($HBSI_g$) with χ=2.84 and for boys ($HBSI_b$) with χ=2.68. Values for χ were obtained from the biomechanical models of children's growth of the group PL_1. The values of each χ were considered representative of normal children's growth because neither obesity nor malnutrition was reported in the population of Polish children 1985-1990. [1]To determine if the US and Polish children growth follow the same growth model, the HBSIs for similar periods have been compared between the two periods and between the nationalities of children using an Anova test at $p<0.05$. In addition, the rates of HBSIs change have been estimated as slopes of the linear regression lines between time (calendar year,Y), with the mean year used as discrete indication of time period.

	US_1	US_2	US_3	PL_1	PL_2
Time period	1971 - 1975	1988 - 1994	1999 - 2002	1985-1990	2003-2006
Girls /Boys(N)	1872/1612	2288/2275	2119/2301	444/403	290/283

Table 1. The number of children (n) from which body mass, height, sex and age data were collected for two comparable time periods in American children (US) and in Polish (PL) children.

Results

The HBSIs indices increased ($p<0.01$) over the studied period of time both in US and Polish children (Table 2, Figure 1). The US children exhibited higher ($p<0.01$)

Mean ± SD	US_1	US_2	US_3	PL_1	PL_2
$HBSI_g$	13.81±2.1	14.89±2.52	15.31±3.47	13.21±1.73	13.52± 2.28
$HBSI_b$	14.01±2.6	15.19±3.09	15.80±3.66	13.74±1.72	14.83±2.52

Table 2 HBMI in girls and boys (g,b- superscript respectively) increased ($p<0.01$) between periods 1 and 2 in American (US) and Polish (PL) children.

values of HBSI than Polish children. The rate of HBSI as a function of the time was much greater in US boys ($HBSI_b$ 0.054Y -92.93), US girls ($HBSI_g$ =0.053Y -73.9) and PL boys $HBSI_b$ = 0.061 Y – 117.6, than in PL girls ($HBSI_g$ =0.019Y - 24).
A steady increase of the HBSI (figure 1) indicates a proportional increases in body width and depth. Such changes in the body distribution are typical for the body composition changes in particular obesity, which has been identified as a major risk factor. [2] HBSI values in Polish children were lower for the studied period of times, however they demonstrated the same rate of increase in Polish boys as in US boys suggesting that the same factors, (with delay in Poland comparing to USA) affect the male samples in both countries, The fact that the HBSI in Polish boys increased much faster than in Polish girls, requires more detailed study. These differences may relate to sex specific traits in the eating habits and life style choices of Polish boys and girls.

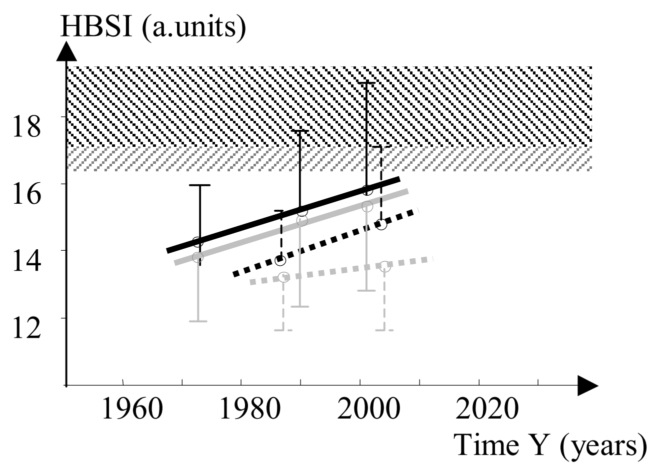

Figure 1. Increase in the HBMI , in girls (gray) and boys (black)in US (solid line) and Polish (dashed line) children as a function of time (Y).

Conclusion

The application of HBSI allows for rapid comparisons between populations of children and it substantially simplifies the analysis of HBS by avoiding tedious growth chart analyses. To account for the changes in the HBS of pediatric subjects who undergo multiple kinemics (including gait) analysis BMIs should be replaced by age invariant HBSIs.

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INTER-JOINT COORDINATION OF YOUNG PEOPLE WITH AUTISM AND ASPERGER'S DISORDER DURING WALKING AT DIFFERENT SPEEDS

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Introduction

Disordered movement such as motor clumsiness and awkwardness constitutes a significant clinical feature of autism and Asperger's disorder (AD). In comparison to normal control groups, specific features of autism include increased variability and inconsistency in movement control during gait whereas in AD, coordination and overall smoothness in gait is reduced[1]. It is hypothesised these differences may indicate greater cerebellar dysfunction in autism than in AD. The aim of this paper is to evaluate whether inter-joint coordination patterns may contribute to quantifying and/or delineating basal ganglia and/or cerebellar motor features in young people with autism and AD during walking.

Statement of clinical significance

Current clinical descriptions of motor dysfunction are imprecise and lack an empirical basis. The elucidation of objective criteria that may help define if autism and AD are distinct disorders with separate neurobiological underpinnings, may enhance current diagnostic methods that rely on subjective assessment. Quantitative gait analysis can provide this objective evidence, which may further inform clinical descriptions.

Methods

The preliminary data set includes a comparison group of normally intelligent children (C) (n=9, 11.0±0.6years, 148.9±5.3cm, 42.1±9.9kg), a group with high functioning autism (HFA) (n=9, 10.5±1.2years, 143.3±15.0cm, 38.6±13.4kg) and a group with AD (n=9, 10.6±1.1years, 144.0±8.4cm, 37.5±5.8kg). Three-dimensional data were collected using an 8-camera motion system. Five trials at self-selected preferred, fast and slow speeds were captured. Anthropometric measurements and marker placement were conducted by one of 3 clinicians to the same standardised protocol. A conventional biomechanical model was used to calculate full body joint kinematics for 2 left and 2 right strides of each trial. Using methods described previously[2], phase portraits, relative phase and deviation phase (DP) were calculated to reveal patterns and stability of inter-joint coordination. One-way ANOVAs with factor Group (C, HFA, AD) were performed to determine significant differences in DP and Bonferroni multiple-comparisons performed to find which groups were significantly different.

Results

Phase portraits indicate increased variability at the ankle during stance in both the HFA and AD groups with the path of the trajectory and variability of the portraits changing with gait speed. An example of an ankle phase portrait at slow speed of an AD participant is demonstrated in Figure 1a. Hip-knee and knee-ankle relative phase plots indicate timing of the reversal in coordination for both the HFA and AD groups is variable during stance, yet the timing and pattern of hip-ankle relative phase is reasonably consistent. An example of a hip-knee relative phase plot at slow speed of the same AD participant is demonstrated in Figure 1b. Significant differences (p<0.05) in DP values were mainly at slow speed and only during swing phase when inter-joint coordination was largely in-phase (Table 1).

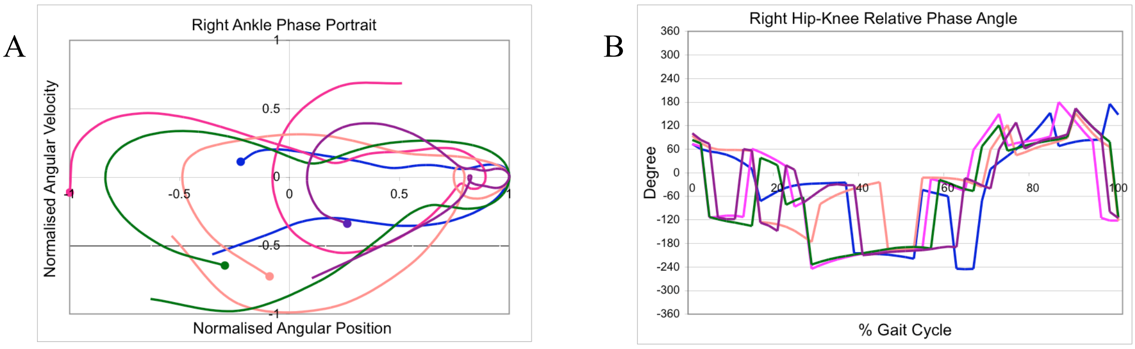

Figure 1: A – Right normalised ankle phase portrait of five stance phase cycles at slow speed of one AD participant. The filled circle represents initial contact. B – Right hip-knee relative phase angle plot of five gait cycles at slow speed of the same AD participant.

DP Value (°)	C		HFA		AD	
Hip-Knee	Left	Right	Left	Right	Left	Right
Preferred	30.4 (5.5)	25.0 (7.2)	34.6 (7.1)	35.0 (8.9)#	30.4 (7.5)	35.1 (6.9)*
Slow	31.5 (8.3)	31.5 (5.3)	41.2 (9.3)	37.6 (10.5)	45.0 (12.3)*	44.6 (12.6)*
Knee-Ankle						
Slow	67.5 (15.2)	62.8 (6.5)	86.3 (14.2)	76.8 (17.9)	81.2 (17.2)	82.5 (19.3)*

Table 1: Significant swing phase DP values with #p<0.05 HFA v's C and *p<0.05 AD v's C

Discussion

The variability in the ankle phase portraits suggest some disorder in the motor system in the HFA and AD groups. The stability of the stance knee is influenced by this variability, resulting in changes in the coordination between the thigh and shank segments, particularly at slow speed in the AD group when instability of the swing limb is at its greatest. Such instabilities may represent basal ganglia motor features such as those seen in Parkinsonian gait[3], which is consistent with the overarching hypothesis that the movement disorder associated with AD is predominantly associated with basal ganglia disruption. Alternatively, the variability in timing in coordination of multiple joints may be suggestive of cerebellar involvement. The small sample size precludes definitive conclusions with further evidence from a larger study sample currently being collected to further clarify group differences.

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SPEED CHANGE OR TURN INITIATION: ANKLE MOMENTS IN TURNING

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INTRODUCTION

Nearly all gait research has studied walking straight ahead at constant speed. However, for walking to be functional humans must also negotiate corners. Turning gait has more frequent falls and more severe injuries than walking straight ahead[1]. The kinematics and kinetics of turning is especially difficult to quantify because people slow down as they approach a turn and speed up after the turn[2]. This has obscured the mechanisms of turning because the ankle kinetics involved in decelerating and accelerating gait[3] mask the ankle kinetics to initiate and terminate turns. This study compares ankle kinetics during turning, accelerating and decelerating to elucidate the kinetic mechanisms of negotiating a frequently-occurring obstacle: a 90° hallway corner at self-selected speeds. This study aims to determine if the sagittal ankle moment is generally used to control speed or to control turn direction.

CLINICAL SIGNIFICANCE

Understanding this common, challenging and poorly understand variant of human gait may lead to more effective interventions to improve functional mobility in real-world settings.

METHODS

Ten normal adults gave informed consent to participate in this study. Each subject walked around a 90° hallway corner, and then four conditions walking straight ahead: Accelerating (1.0 to 1.4 m/s); Decelerating (1.4 to 1.0 m/s) and Fast (1.4 m/s) and Slow (1.0 m/s). Full-body gait kinematics and kinetics were collected on ten trials in each condition using a 12 camera Vicon MX System and Plug-In Gait model (Oxford Metrics, Oxford, UK). Force plate contacts were recorded for each trial. Sagittal ankle moments at 25% of the gait cycle were compared across conditions. An ANOVA, with Scheffe's tests posthoc, were used to test hypotheses; p set at 0.01 (Bonferroni correction.)

RESULTS

Foot contacts around the 90° hallway corner were categorized as 1) Initiating the turn; 2) the Apex of the turn; 3) Terminating the turn (see figure). As expected, the Turn Initiation Step, the Apex Step and the Deceleration Step were each significantly different from the Turn Termination Step and the Straight Acceleration Step ($p < 0.001$). The Turn Initiation Step, the Turn Apex Step and the Straight Deceleration Step were not significantly different from each other ($p > 0.79$-0.99). The Turn Termination Step and Acceleration Step were also not significantly different from each other ($p > 0.90$). The Fast Slow Steps were not statistically different from each other ($p > 0.98$).

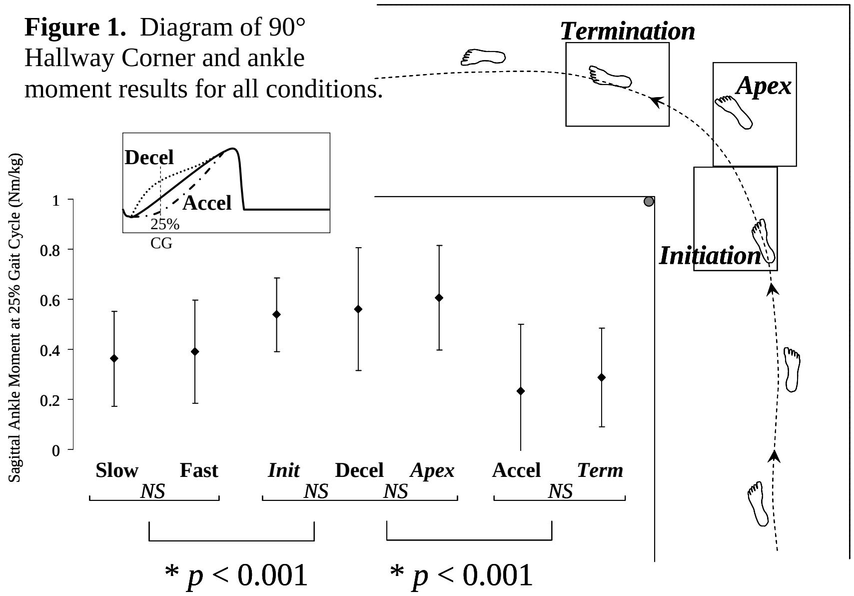

Figure 1. Diagram of 90° Hallway Corner and ankle moment results for all conditions.

DISCUSSION

These data support the hypothesis that the mechanisms of deceleration and acceleration used during turning in a 90° hallway corner are principally for gait speed control and do not necessarily initiate, continue or end the turn by applying off-center accelerations to alter the straight-ahead path of the individual. Therefore, humans do not appear to accelerate or decelerate one side of the body as the preferred strategy to accomplish turn. It is more likely that foot placement alters the mediolateral shear forces which accelerate the body's center of mass in the direction of the turn[2], and that the rotation moment of the hip, knee and ankle control the orientation of the trunk to keep it facing the direction of travel. These subtle moments, which have a sinusoidal pattern of internal and then external rotator moments as stance progresses, are thought to provide the rotational control of the trunk primarily in late stance, like the rudder on a boat [4]. Small errors in these rotator moments in late stance could account for the eight-fold hip fracture injury risk observed during turning [1]. If errors in rotator moments fail to rotate the trunk to face the direction of the turn, the fall would essentially be lateral, making the greater trochanter the initial contact point of the body with the ground and increasing injury risk. Errors in foot placement and rotator moments may be the principal mechanism of hip fracture during falling, and not tripping over an object, where the body generally falls forward. Glaister, et al,[5] has shown that turning steps account for 30-50% of all steps during typical indoor walking behaviors, but successful turning is not generally evaluated in clinical gait analysis, or visually in a clinic hallway or a goal of rehabilitation in the elderly, amputees or stroke survivors. Perhaps an increased understanding of the mechanisms of turning will enhance our ability to intervene and improve performance on this functional task necessary for everyday mobility.

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The effect of water exercise on static and dynamic balance in older women

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Introduction: Risk of acute and chronic diseases increase with advancing age; strength and functional abilities decrease with aging. The changes occur in functioning of skeletal- muscle, somatosensory, visual and vestibular systems involved in balance and postural control. Impaired balance is one of the most commonly identified risk factors for falling and thus subsequent injury (15). Typical strategies to improve balance and mobility include exercise programs (6, 7, 9, 10). One challenge in developing and designing such programs is identifying a safe medium coupled with an effective program that can be used to enhance balance. Recently, the use of water exercise has become wide spread in rehabilitation and fitness maintenance (13). Because of its low impact and reduced risk nature, water exercise has been suggested as a safe medium for older individuals (13). However very little are known about the effects of this type of exercise on balance in older individuals. We hypothesized that participants in a water exercise program should demonstrate better balance than those who didn't take part in such a program.

Method: 30 participants, mean age 62±3/1 years, were recruited from aging clubs in Tehran. Participants were randomly assigned in control and training groups. Exercise training was performed as a group activity in water, for 60 min, 2 days per week in 6 weeks. Postural sway parameters, including average displacement and velocity of center of pressures (COP) in the ML and AP directions in the one leg stance position, as a measure of static balance and functional reach forward and lateral left and right as measures of dynamic balance were measured in both groups. Standing balance on the dominate leg was determined using a force plate (BERTEC 40×60), with a sampling frequency of 50 Hz. The data were smoothed using a Butterworth filter with a cutoff frequency of 12.5 Hz. The COP velocity was calculated using the following formula (5):

$$\frac{f}{n-1}\sum_{i=1}^{n-1}\sqrt{(cop_{(i-1)}-cop_{(i)})}^{2}$$

Where n is number of data samples, f is the sampling frequency and cop is the displacement of the cop in the ML and AP directions. Functional reach was determined while participants were asked to reach forward or lateral extends at a fixed base of support in standing position (3). Paired t – tests were also performed for exercise participants and the control group before and after the training program at significant α level of 0.5.

Results: Table 2 shows the results of the comparisons between the two groups before and after the training program. Compared with the controls, exercise subjects demonstrated statistically significant differences at retest in FRT, FRTC and cop and cop velocity in the ML direction.

Table 2: Mean and SD of static and dynamic balance measures

Group	Time	FRLT (cm)	FRRT (cm)	FRT (cm)	COP VEL AP (Cm/s)	COP VEL ML (Cm/s)	COP AP (cm)	COP ML (cm)
Control (n=15)	Pre-test	22.6 ±7.4	20.8 ±6.9	24.9 ±5.5	5.1 ±5	3.08 ±4	4.1 ±٠/5	2.08 ±./4
	Post-test	24.9** 5.5	22.9** ±3.5	26.3** ±2.4	4.9 ±1.25	4.9** ±2.5	3.89** ±1.35	3.1** ±5.5
Exercise (n=15)	Pre-test	*23.7 ±6.8	22.1 ±7.6	*25.2 ±6.4	3 ±5	*3.8 ±1.4	2.6 ±٠/5	*2.9 ±٠/4
	Post-test	28.8 ±5.7	23.9 ±5.1	29.9 ±4.2	2.9 ±/.95.	2.45 ±1.1	3.80 ±1.30	1.89 ±./45

❊ P ≤ 0/05: Significant differences before and after training

❊ ❊ P≤ 0/05: Significant differences before and after training

Discussion: The results of this study demonstrate that the use of water exercise can produce improved postural sway and functional reach in older adults. The results of this study suggest that one leg stance improves with training, in the ML direction. These results are in consistent with those of Nagy (2007), Nagy (2004), Rogers (2004) and McGleagan (1995). It seems that balance in ML direction, in comparison with AP direction is more sensitive to the effects of training (8). In the AP direction as compared with the ML direction, there are an increased number of alternative strategies that the individuals can use to cope with instability (8). Reasonably, physical activity prepare appropriate and more challenges for balance mechanics and improving them (9).The findings of this study conflict with the conclusions of Stones and Kozma(1987).

We detected a significant improvement in FRT after training, that is in agreement with findings of Douris(1996), Simmones and Hanson(1996) but in conflict with bellew (1996). Regarding the effects of water exercise on balance, results of this study are in agreement with Doris (1996), Simmons & Hanson (1993). Combination of frequency and speed of movements might lead to enhanced strength (15), endurance (7), flexibility (6) and reaction time. Balance, strength and proprioception may be addressed in an aquatic environment (12). Routi and associates described the support offered by water as allowing more independent upright posture.

Conclusion: In conclusion, these findings indicate that combined training with an emphasis on senses of balance can be more effective than programs that stress only balance, flexibility or aerobic training. Also it is possible that multi senses training that manipulate systems involved in balance control in stable and non stable base of support, to be effective to improve balance in older adults. The use of water exercise because of its safe nature and challenging of balance systems can be an effective method to improve static & dynamic balance, particularly in the ML direction.

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Improve Rehab Patient Care with Laban Specification and Wireless Sensor Tracking

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Introduction

In phase 2 of International Cooperation on Human Body Movement Analysis Project, a Department of Dance, an Academic Department of Computer Science, and a Clinical Motion Laboratory worked together as a team on research to improve rehab patient care by applying Labanotation, computer modeling/animation techniques, and wireless sensor tracking systems for specifying and recording rehab exercises. We have identified four key rehab exercises and produce the related high speed high resolution video and related mocap data for further study. We have investigated the use of Labanotation for specifying these key rehab exercises due to its precise and concise notation and because the ease of recording and editing its notation[1], and proposed the notation enhancement or customization specific for rehab exercise specification. LabanDancer software was assessed and its usage as a module for a rehab exercise specification, animation, tracking, and analysis system was evaluated. We proposed a wireless sensor based tracking system architecture for improving rehab patient care. Research is conducted to understand the limitation of Cricket based mica2 wireless sensor system. A utility program is developed to take C3D data as input and efficiently output joint locations, which can be displayed by Maya as desired.

Statement of Clinical Significance

We explore the use of Labanotation for the key rehab exercises and discuss the enhancements needed for the rehab exercise specification. The reason for these enhancements is not really the inadequate of the Labanotation. It is more the result of different emphases between dance specification and rehab specification. In rehab exercise specification, it emphasizes more on the precision such as "Raise the leg out to the side for about 12 inches" or on relative human terms such as "Place feet shoulder width apart." But overall we were impressed with how succinct and precise the Labanotation can be used to describe these exercises. While traditional Labanotation has been used in dance and movement observation, its use "for representing movement that occurs in movement-based interaction with technology"[2] has been studied.

Methods

A team consisting of experts on gait and motion analysis, dancing and Laban motion analysis, and physical therapy, identified a set of four key rehab exercises for this project. The selected rehab exercises are videotaped with the brand new high end high resolution Vicon MX camera system in the Gait and Motion Analysis Lab at The Children's Hospital, Denver, with four different camera angles and the related motion captured data are saved in C3D data format for further study. These valuable data are used as a reference data set for the design and implementation of computer animation and human motion tracking systems. The four key rehab exercises include mini-squats, posterior pelvic tilt, shoulder elevation with rotation, and standing hip abduction.

Discussion

Figure 1 shows a proposed system for rehab exercises and how can be integrated. The system will consist of a set of rehab exercises database with exercises specified precisely with Labanotation and augmented with safe zone information which indicates the safe boundary of rehab motion. The sensor tracking subsystem can be attached to collect the patient rehab exercises using low cost ultrasound based wireless sensors. A joint calculation utility can be used to analyze the patient sensor tracking data and calibrate the size of the patient. The Laban model and animation module can read the rehab exercise in Labanotation and present 3D animation sequence on screen to help training and provide guidance of the exercise. The C3D data generated by the Laban animation model and the patient sensor C3D data can be analyzed at the Rehab/C3D sequence comparison module and produces alerts or guidance information when the patient movement is out of the safe zone or too slow. A side-by-side comparison of animation sequence and patient rehab movement will be a great enhancement for the rehab patients and physical therapists alike.

Results

We have produced a set of four valuable key rehab exercises with rather complete MPEG2 video, mocap C3D data, related Labanotation specification for these exercises. These data can be used as a basis for further study or for evaluating the computer animation system and sensor based tracking system. It also provides an excellent example for researcher to discuss how a new generation of human motion specification or the current Labanotation can be enhanced. A system's architecture is proposed where rehab exercises can be specified, demonstrated with 3D animation, recorded with low-cost wireless sensors. As a telemedicine application of the system, the recorded exercise data can be transferred to physical therapy experts for evaluation and new or modified exercises can be downloaded remotely. We have identified the challenges in the design of such system and believe the prototype if commercialized will greatly improve the rehab patient care.

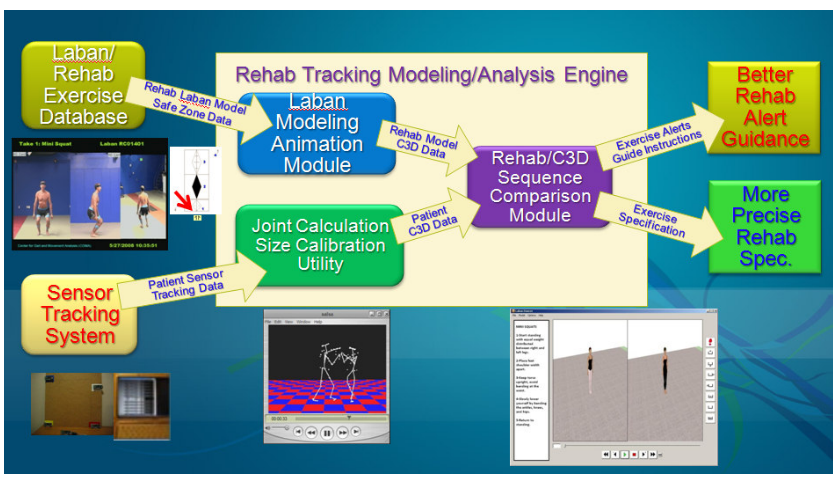

Figure 1. HMTR System's Flow.

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A CUSTOMIZED APPROACH TO PARTIAL BODY WEIGHT SUPPORT REHABILITATION IN PATIENTS WITH POST-STROKE HEMIPARESIS

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INTRODUCTION

For many individuals with different forms of gait ability limitations, treadmill training in conjunction with partial body weight support (PBWS) has proven an effective rehabilitation strategy with regard to the restoration of gait and the functional transfer to overground walking velocity [1-3]. Despite the established effectiveness of PBWS rehabilitation, support schemes intended for the functional deficits of individual gait impairments are limited. For example, in patients with post-stroke hemiparesis, Hesse et al. (1999) concluded that with passive support, patients utilized the support force almost exclusively during paretic limb single support, implying that excessive unloading of the healthy limb is unnecessary and may result in unnatural gait pattern retraining. There is an apparent need for gait synchronized support modulation exclusively intended for unilateral walking limitations as can only be achieved through an active feedback controlled PBWS system [4]. Here we present the ability to provide both constant 30%BW support and two gait synchronized unilateral support profiles, and examine the affect of each on the vertical ground reaction forces, temporal-spatial characteristics, and center of mass (CoM) height in three patients with post-stroke hemiparesis.

CLINICAL SIGNIFICANCE

To provide optimal rehabilitation, PBWS should promote efficient lower extremity movement patterns [1-3], and should allow natural sinusoidal CoM motion which is critical to efficient and comfortable gait [5]. In the case of post-stroke hemiparesis, modulated support can allow for customized support patterns which promote appropriate CoM movement, and provide desired unloading of the paretic limb without unnecessary unloading of the unaffected limb in hopes to restore normal symmetrical walking ability. We believe that the use of our actively controlled PBWS system [4] can contribute to developing more optimal and clinically effective rehabilitation protocols.

METHODS

The structure of and the feedback control algorithms utilized by our PBWS system have been explained in detail elsewhere [4]. Study subjects were three females of ages 36, 43, and 79 yrs with hemiplegic gait as a result of cerebrovascular accidents 12, 21, and 50 months prior to the study, respectively. Patients had completed a course of inpatient rehabilitation followed by outpatient therapies including exposure to other PBWS devices. Kinematic data from the torso and lower extremity were obtained using a 10 camera Vicon 624 motion analysis system (ViconPeak, Lake Forest, CA, USA). Corresponding kinetic data were captured using an instrumented treadmill set to the patient's unassisted overground walking velocity. Unsupported walking and a constant 30%BW support were examined in addition to two gait synchronized modulated support conditions: 30-0MOD (30%BW support during paretic limb stance transitioned to 0%BW during healthy limb stance); and 30-15MOD (30%BW support during paretic limb stance transitioned to 15%BW during healthy limb stance). Six consecutive cycles of gait were averaged for the healthy and paretic limbs for each condition. Average curves were prepared for the healthy and paretic vertical GRF components and the CoM height. The magnitude and asymmetry [SI=$|X_R-X_L|$/average($|X_R|$,$|X_L|$)×100] of stride length, cadence, and single limb support duration were also examined.

RESULTS

Each of the three subject's vertical GRF comparisons is shown in Figure 1. By design, all three conditions reduced the paretic limb vertical GRF by approximately 30%BW. The use of gait synchronized support allowed for customized unloading of the healthy limb while maintaining desired unloading of the paretic limb. Constant support exaggerated the vertical CoM displacement and/or caused this displacement to be much more asymmetric. 30-15MOD minimized excessive vertical CoM displacement and best promoted CoM symmetry. For all conditions, asymmetry between the paretic and healthy limbs was less than 2.2% for both stride length and cadence, and thus the right and left average was examined. Beneficial decreases in cadence and increases in stride length were observed for constant support (3/3 subjects), 30-0MOD (2/3 subjects), and 30-15MOD (3/3 subjects) as compared to unsupported walking. Considerable asymmetry was observed in single limb support duration during unsupported walking (mean, 15.7%) which was noticeably increased for all subjects during constant support (mean, 23.2%). 30-0MOD tended to promote a more symmetric single limb support duration (2/3 subjects).

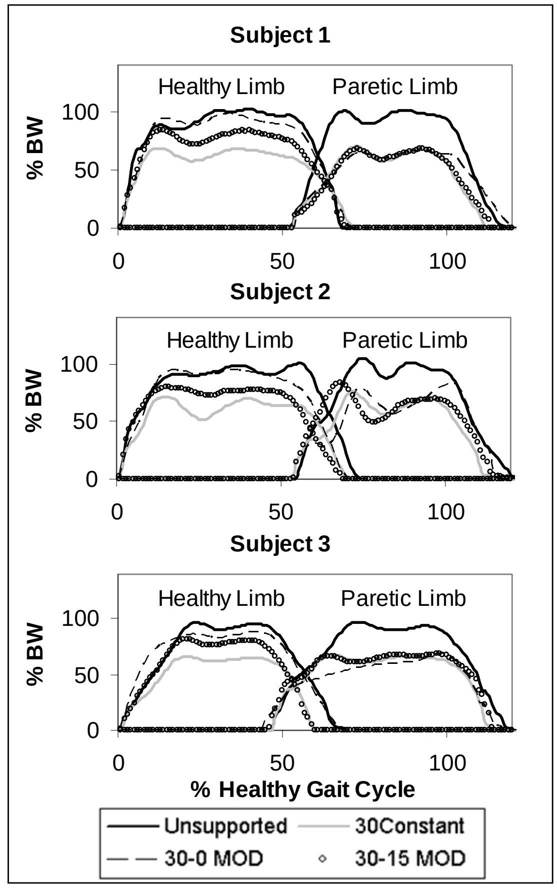

Figure 1. Vertical ground reaction forces of the healthy and paretic limbs for each condition.

DISCUSSION

Although no conclusions can be drawn regarding the long-term clinical implications of the presented PBWS interventions, the ability to provide gait synchronized support may allow clinicians the potential to manipulate support characteristics to optimize gait pattern retraining in a customized, patient-specific manner.

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Rehabilitation of physical injuries in young dancers

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Background: In the pursuit of excellence and self-accomplishment through the physical practices of dance, the dancer continually faces the challenge of dealing with injuries.

Purpose: To minimize the risk of injury in young non-professional female dancers and to rehabilitate the injured dancers.

Methods: 1336 female dancers, age 8-16 years were screened. 61 different types of injuries and symptoms were recorded and later classified into four major categories (knee injuries, foot and ankle injuries, non-categorized injuries and low back pain). The risk factors considered for injuries were: joint range of motion (ROM), body structure, anatomical anomalies and dance discipline.

Results: Hypo or hyper joint ROM as well as the presence of scoliosis, significantly increase the risk for injury among young dancers. Hours of practice per week increase the risk for injury only after the age of 13. BMI was not found to be significantly associated with injury.

Conclusions: To prevent injuries and to rehabilitate the injured dancers, dancers should be aware of their joint ROM limitation and avoid any efforts to compensate for it in other joints. Slight stretching exercise of rigid soft tissues as well as strengthening exercise for less stable joints (to improve muscle function) is essential for decreasing chances of injury. As dancers with scoliosis are at higher risk to develop LBP, all dancers must be screened for anatomical anomalies and be aware of limitations imposed by their physique.

Changes in upper extremity weight bearing over time for children diagnosed with CP

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Introduction:

To date, most literature on the gait pattern of children diagnosed with cerebral palsy (CP) has focused on lower extremity kinematics and kinetics. A limited amount of data has been reported on the motions of the upper extremities of children diagnosed with CP. Strifling et al (2008) reported similar patterns of upper extremity motions for 10 children diagnosed with CP when walking with a forward and reverse walker.(1) Fast et al (1995) reported that adults with decreased balance due to upper motor neuron involvement demonstrated 34 % of upper extremity weight bearing through a walker.(2) To date, the amount of change in upper extremity weight bearing while walking has not been reported for children diagnosed with CP.

When using a walker there is difficulty tracking and identifying surface passive markers during three-dimensional motion analysis studies and often an inability to obtain lower or upper extremity kinetic data. To combat this problem, in 1996, we designed and built a "universal walker" which would adjust to children and adolescents of all sizes and ages. In each hand grip is an AMTI six degree of freedom transducer which allows us to obtain upper extremity weight bearing as a percent of body weight for each arm.

Clinical significance:

By reviewing our database an assessment of the changes in upper extremity weight bearing could be assessed. To date no studies have reported the changes over time in upper extremity weight bearing for children diagnosed with CP.

Methods:

The study was IRB approved. The universal walker weighs 42 lbs (19 kg) and requires just 2 lbs (0.9 kg) of force to move it in a forward or backward direction. Since 1996 the universal walker has been used on children between the ages of 3 to 18 years old. The children have been of various heights (99 cm to 188 cm tall) and weights (18 kg to 132 kg).

The purpose of this study was to answer the following questions:

- What are the characteristics of upper extremity weight bearing for children diagnosed with CP?
- What are the changes in upper extremity weight bearing for children diagnosed with CP over time after undergoing surgical intervention?
- What are the changes in upper extremity weight bearing for children diagnosed with CP without undergoing surgical intervention?

Repeated measure ANOVA and post hoc test between groups (different orthopaedic interventions, and no interventions) were performed using SPSS 15.0 (Chicago, IL).

Results:
A retrospective review of our database revealed 1049 separate gait studies were performed using the universal walker. Of these studies 192 subjects used our universal walker in their first gait study. After eliminating subjects without a second study, or subjects without complete datasets, 62 children demonstrated complete datasets for two studies performed less than 3 years apart. Twenty-four children had no intervention between the first and second study. Thirty-eight of the children underwent an orthopaedic intervention between the first and second study.

There were no statistical differences between all groups for age, height, weight, or BMI at the time of the first study ($p>.05$).

There were no statistical differences between all groups based on interventions for temporal spatial data between the first and second study ($p>.05$). However, there was a statistically significant increase in upper extremity weight bearing between the first and second study from 14 % (95 % CI 12-18) to 18 % (95 % CI 16-20) of body weight through each upper extremity ($p<.05$) for all groups. Interestingly, no statistically significant differences were noted between groups of subjects based on interventions ($p>.05$).

From the first to the second study, some children demonstrated a large decrease in upper extremity weight bearing through each upper extremity (from 40 % to 18 % body weight); while others demonstrated a large increase in upper extremity weight bearing through each upper extremity (from 11 % to 35 % of body weight). An analysis for correlations between patient demographics and changes in upper extremity weight bearing was performed. Significant correlations were noted between the change in upper extremity weight bearing and the following measures: age ($r = -0.62$), BMI ($r = -0.55$) and amount of upper extremity weight bearing at the time of the first study ($r = -0.74$).

Discussion:
On average an increase in upper extremity weight bearing was noted between the first and second study for all subjects. We currently use the information obtained from our universal walker (the percent of body weight supported through the upper extremities) as one indicator for consideration of performing surgical intervention. The more upper extremity weight bearing reported the less likely surgical intervention will be recommended.

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Upper Extremity Motion Analysis in Tetraplegia:
Quantitative Outcomes after Tendon Transfer Surgery to Improve Function

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Patient History

A 14-year-old female (MM) with C6 ASIA A tetraplegia secondary to a cervical spinal cord injury sustained in a motor vehicle accident 15 months prior was referred to the Motion Analysis Laboratory (MAL) for evaluation to assist with surgical planning. She had sustained a C4 pedicle fracture and compression fractures of C4-5, with subsequent surgical fusion of C4-C7. Tendon transfers for improved pinch and elbow extension strength were being considered prior to her first MAL Evaluation. She was evaluated on three separate occasions in the MAL, twice before her first surgery. She then underwent orthopaedic surgery on her right upper limb, and subsequently her 3rd MAL evaluation documented post-operative outcomes for her right upper limb before identical surgical procedures on the left upper limb.

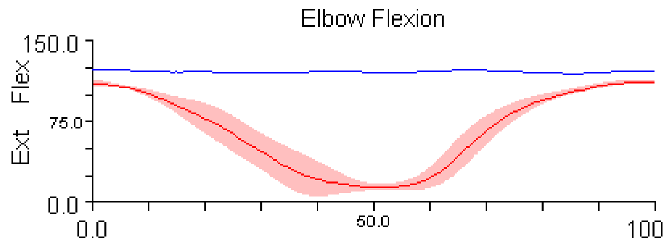
Clinical Data

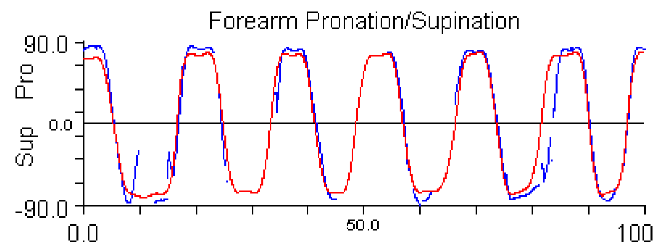
Pre-operative musculoskeletal examination findings were consistent with C6 tetraplegia. Proximal shoulder girdle atrophy of triceps (posteriorly) and pectoralis minor (anteriorly) was apparent bilaterally. Shoulder motion and strength were otherwise good throughout. At the elbows, all flexors were strong with 5/5 strength in biceps, brachialis and brachioradialis. Elbow extensors had 1+/5 strength with only 10° active extension against gravity from a fully flexed position, thus completely non-functional. Forearm supinators had 5/5 strength. Forearm pronation was accomplished only by brachioradialis initiating pronation and then full pronation occurred using gravity and proximal positioning of the arms in space. Pronator teres was 1/5 strength; pronator quadratus was 0/5. Wrist extensors were strong (5/5); extensor carpi radialis (ECR) (C5-8 innervation) was a bit stronger than extensor carpi ulnaris (ECU) (C6-8). Wrist flexor strength of flexor carpi ulnaris (FCU) and flexor carpi radialis (FCR) was absent (0/5). Remaining hand and finger muscle strength was absent throughout with the following exceptions: flickers of movement (2-/5 strength) were noted in left flexor digitorum superficialis (FDS) to middle and ring fingers, abductor pollicus longus (APL), abductor pollicis brevis (APB), opponens pollicis and opponens digiti minimi. Metacarpophalangeal (MCP) thumb joints were hypermobile in all directions (lax joint capsules). Collateral and volar ligaments were stable bilaterally. Atrophy was noted in thenar and hypothenar eminences bilaterally. Grip strength using a Jamar dynamometer was 0 lbs. on both right and left sides. Three-jaw (palmar) pinch strength was ¾ lb. on right and left sides. Lateral (key) pinch strength was ¼ lb. on right and left sides. Occasional spasms occurred in left triceps, which triggered spasms in the left FCU. Selective motor control was excellent, enabling finger dexterity and skilled grasp/release of only very light objects when using tenodesis action driven by the strong wrist extensors and passive tension in thumb/finger flexor tendons. The Jebsen-Taylor Test of Hand Function and a Functional Hand Grasp Assessment were also performed.

Treatment Indications & Decisions

MM's goals included anti-gravity movement and increased strength/power of elbow extensors, forearm pronators, wrist flexors, and pinch strength to perform functional tasks such as keyboarding, applying her own make-up, picking up objects weighing 1+ lbs., opening doors, self-transfers, writing, improved dressing skills, cutting food with a knife, and improved cosmesis. In an attempt to address her goals, the decision was made to perform staged procedures (right upper limb first) to improve anti-gravity elbow extension and lateral pinch strength. Surgical procedures included biceps to triceps tendon transfer (medial approach), and brachioradialis to flexor pollicis longus (FPL) tendon transfer with splitting of the FPL tendon at the interphalangeal (IP) joint of the thumb for stabilization of this joint during lateral pinch.

Motion Analysis Data *(Pre/Post)*
Anti-gravity elbow extension with right elbow post-op (mean of 5 trials with ± 1 SD band) and left elbow ("flat line") pre-operative data are depicted in the first graph. In the second graph, individual trials of right forearm pronation/ supination are shown illustrating the nearly 180° arc of post-operative motion. Visual 3D software and a customized six degrees of freedom upper extremity kinematic model were used for data collection and reduction. (See 1st reference.)





Outcomes/Summary
After the biceps to triceps tendon transfer on her right side, MM gained the ability to extend her elbow fully against gravity, resisting 5.5 pounds of pressure as measured with a dynamometer, with only a very minor loss of elbow flexor strength. She improved her transfer abilities. Eventually she became able to bench press 8 lbs. Since the brachioradialis to FPL tendon transfer on her right, MM is now able to hold numerous items in her right hand with 3 lbs. of lateral pinch strength, compared with pre-operative abilities where she could only grasp items using tenodesis without any power at all (only ¼ lb. of lateral pinch strength). Additional skills gained since surgery *that were not anticipated* included the ability to flex her right wrist against gravity, as well as pronate her right forearm against gravity and take resistance. These abilities became possible because the brachioradialis to FPL transfer now crosses the wrist joint *enabling anti-gravity wrist flexion*, as well as inserts radially into FPL *enabling anti-gravity pronation* with contraction of brachioradialis. Quantitative documentation of these outcomes was possible using MAL data, adhering to a specific anti-gravity movement protocol. A sophisticated upper extremity kinematic model enabled accurate measurement of anatomic joint movements.

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INFLUENCE OF WHEELCHAIR BOUNDARY ON BREATHING PATTERN IN LIMB-GIRDLE MUSCULAR DISTROPHY

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Introduction

Limb-Girdle Muscular Dystrophy (LGMD) was first described as clinical entity in 1954 but only in the 1990s different genes and proteins were associated to the different diseases [1]. With molecular clarification, the need to recognize characteristic patterns of disease through the clinical assessment is emerged. Respiratory muscle weakness resulting in symptomatic hypoventilation and respiratory failure is found in a few of the LGMD. No detailed data concerning respiratory function in LGMD are reported, as long as the pulmonary and cardiac functions seem to be preserved in these diseases. By using spirometry and Optoelectronic Plethysmography (OEP, [2]) we described breathing pattern of LGMD and we investigated differences between LGMD ambulant and wheelchair bound patients.

Materials and Methods

27 adult patients affected by LGMD and 20 healthy age and sex matched volunteers (Control Group – CG) were involved in this study (Table 1).

Table 1. Pathological and healthy groups' data. (A: Ambulant, W: Wheelchair bound)

	N	Sex	Age (years)	Weight (kg)	Height (cm)	Walking Ability
LGMD	27	11 F, 16 M	35.2±14.7	63.3±15.8	169.4±8.2	15 A, 12 W
CG	20	6 F, 14 M	32±9.2	70.9±14.8	174.9±7.6	

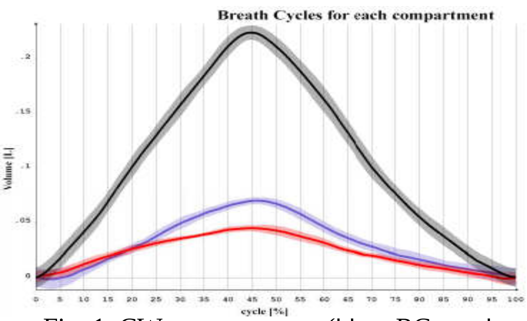

Fig. 1. CW compartments (blue: RCp, red: RCa, grey: AB) during QB in SU position for an healthy subject

Patients underwent spirometry and OEP. OEP technique allows a quantitative description of breathing pattern; it's based on the measurement by an optoelectronic system (OEPSYSTEM, Bts Bioengineering , Italy) of 3D motion of reflective markers positioned on subject's trunk and on an algorithm based on Gauss' theorem that provides the measurement of chest wall (CW) volume and its thoraco-abdominal compartments (RC_p: Pulmonary Rib Cage, RC_a: Abdominal Rib Cage, AB: Abdomen). Breathing pattern evaluation was performed in Seated (SE) and Supine (SU) position and in both the conditions the subjects were asked to quietly and spontaneously breath for three minutes. Starting from CW volume traces, several consecutive breaths were chosen to obtain a single representative normalized breath. For each subject the complete ventilatory pattern for SE and SU conditions was defined and Minute Ventilation (V'_E, [L/min]), Breathing Frequency (RR, [min^{-1}]), Tidal Volume (V_T, [L]) and CW compartments percentages ($\%RC_p$, $\%RC_a$, %AB) were calculated (Fig. 1). $\%RC_p$ depends on the activity on pleural pressure and expiratory and inspiratory rib cage muscles, $\%RC_a$ on the abdominal pressure and %AB revealed the activity of diaphragm and the abdominal expiratory muscle. For the LGMD breathing pattern characterization, mean and standard deviation for all the parameters were calculated and then compared with CG. For the analysis of the wheelchair boundary effect, pathological subjects were divided into two groups according to their walking ability (Ambulant: LGMD-A, Wheelchair bound: LGMD-W, Table 1). The two sub groups revealed to be homogeneous in age and sex.

Results

Spirometry data showed a mild involvement in respiratory function, especially for LGMD-W. Between the two subgroups significant differences were found for all the parameters except for FEV1/FVC and RV, with a more significant respiratory involvement in LGMD-W.

Table 1. Spirometry data (% of predicted values, m: mean and st.d: standard deviation) for pathological subjects

	FVC		**FEV1**		**FEV1/FVC**		**PEF**		**TLC**		**VC**		**RV**	
	m	st.d	**m**	st.d	**m**	st.d	**m**	st.d	**m**	st.d	**m**	st.d	**m**	st.d
LGMD	**76.6**	21.8	**78.3**	21.2	**86.4**	6.7	**72**	17.5	**84.7**	19	**76.5**	21.9	**109.7**	34.6
LGMD-A	**<u>88.6</u>**	17	**<u>90.5</u>**	15.7	**86.8**	6.3	**<u>78.2</u>**	16.7	**<u>94.1</u>**	11.8	**<u>87.5</u>**	17.1	**113.2**	29.3
LGMD-W	**<u>61.9</u>**	18.1	**<u>62.9</u>**	16.9	**86**	7.5	**<u>64.3</u>**	15.9	**<u>72.9</u>**	19.9	**<u>62.7</u>**	19.6	**105.4**	41.3

The comparison between underlined data was statistically significant ($p<0.05$)

As concern OEP data, both in SE (Table 2) and in SU (Table 3) position, no differences in ventilatory parameters and CW compartments percentages between the entire LGMD group and CG were found. The comparison between the two pathological sub groups revealed a statistically significant higher AB percentage for LGMD-A in respect to LGMD-W patients.

Table 2. Data of Seated (SE) position

	V'_E		**RR**		**V_T**		**%RC_p**		**%RC_a**		**%AB**	
	mean	st.d	**mean**	st.d	**mean**	st.d	**mean**	st.d	**mean**	st.d	**mean**	st.d
LGMD	**7.4**	2.5	**17.85**	5.23	**0.4**	0.1	**40.60**	14.99	**21.14**	5.75	**37.84**	15.90
CG	**9**	2.3	**15.39**	4.58	**0.6**	0.2	**40.80**	11.16	**23.65**	5.72	**35.55**	10.49
LGMD-A	**7.7**	2.8	**15.78**	4.28	**0.5**	0.1	**36.39**	12.38	**18.99**	5.60	**<u>43.91</u>**	11.89
LGMD-W	**6.7**	1.5	**19.73**	5.47	**0.35**	0.1	**49.25**	17.27	**23.82**	5.06	**<u>26.93</u>**	18.55

The comparison between underlined data was statistically significant ($p<0.05$)

Table 3. Data of Supine (SU) position

	V'_E		**RR**		**V_T**		**%RC_p**		**%RC_a**		**%AB**	
	mean	st.d	**mean**	st.d	**mean**	st.d	**mean**	st.d	**mean**	st.d	**mean**	st.d
LGMD	**6.3**	1.2	**16.91**	5.04	**0.4**	0.1	**23.88**	18.57	**12.99**	7.40	**63.18**	22.22
CG	**5.9**	1.2	**15.37**	5.28	**0.4**	0.2	**21.31**	19.43	**15.14**	2.90	**63.54**	11.28
LGMD-A	**5.8**	1.2	**14.93**	3.99	**0.4**	0.1	**16.06**	13.77	**11.85**	5.82	**<u>72.02</u>**	16.19
LGMD-W	**6.3**	1.5	**19.38**	5.26	**0.3**	0.1	**33.66**	19.64	**14.40**	9.07	**<u>52.12</u>**	24.35

The comparison between underlined data was statistically significant ($p<0.05$)

Discussion

Spirometry data demonstrated that LGMD patients present a mild involvement of respiratory function. An abnormal reduction of FVC, FEV1 and TLC and increased RV, with normal values for FEV1/FVC is typical of restrictive respiratory pattern, probably due to respiratory muscle weakness, particularly evident for non ambulant LGMD subjects. OEP data showed that LGMD patients that are able to walk independently present a breathing pattern similar to that obtained for CG both in seated and supine position. For LGMD patients that were confined on a wheelchair, the diaphragm activity seemed to be less effective and in order to obtain normal ventilatory parameters these patients needed to increase the activity of inspiratory and expiratory rib cage muscles.

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Familiarisation to treadmill walking with body weight support for patients post-stroke. A preliminary report

(data collection is ongoing and more substantial results will be presented at the GCMAS conference)

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Introduction

Various systems have been introduced to support the weight of the body and to prevent patients from falling during gait rehabilitation. One such system termed body weight supported treadmill training (BWSTT) allows treadmill walking while suspending part of the body weight by sling systems and a harness, and is used for many clinical populations including patients post-stroke. Walking with body weight support (BWS) may require learning a new task that one has to be familiarised to [1]. It is not well documented how much practice is needed to achieve steady-state treadmill walking with BWS in patients or if such gait is similar to walking overground. The objective of this study is therefore to investigate if steady-state gait during or following familiarisation can be identified and whether this gait approaches overground gait in ambulant patients post-stroke.

Clinical Significance

Knowledge of the familiarisation process in patients post-stroke, who may have limited capacity for longer training periods, is important for designing clinical trials as well as for the use of BWSTT in the clinical practice.

Methods

Subjects: 13 patients post-stroke with hemi-paretic gait categorised in Functional Ambulation Category (FAC) between 3 and 5, and with preferred walking velocities of 0.51-1.08 m/s have been included so far. The sample consists of 9 males and 4 females aged 29-80 years (mean 58 years) of which 8 subjects have previously experienced treadmill walking. The study is conducted in conformity with the Declaration of Helsinki. The protocol has been approved by the local Science Ethical Committee.

Instrumentation: The subjects walked on a treadmill with a BWS system including harness, pulleys and spring weights. Trunk acceleration was measured using a kinematic sensor attached to the lower trunk collecting data along the anteroposterior (AP), mediolateral (ML) and vertical (V) axes. Results reported here include trunk acceleration amplitudes (RMS), cadence and interstride trunk regularity (autocorrelation coefficients) at preferred velocity.

*Procedure: S*ubjects walked a distance of 7 meters overground, twice at preferred velocity, twice slowly, and twice as fast as they could safely do. Each person then walked 5 minutes on the treadmill for familiarisation at the same preferred velocity as overground, and with 20 per cent BWS. 10 intervals starting every half minute were analysed. All subjects had a 5 minute resting period after the familiarisation session, and were then randomised for order to 20 and 40 per cent BWS sessions at velocities equal to overground walking. Only data for the 10 familiarisation intervals and the test session at 20 per cent BWS at preferred velocity are reported here.

Data analysis: Steady-state gait during the familiarisation period was investigated by examining intraclass correlation coefficients (ICC(2.1)) of variables between consecutive intervals 1-10 during familiarisation, and between the last familiarisation interval (interval 10) and the 20 per cent BWS test period at preferred velocity that followed. This interval was either in the seventh or tenth minute afterwards, depending on the randomised order of testing. Mean ICCs for all variables per interval were calculated using Fisher z-transformed coefficients which were retransformed to obtained means. To explore if treadmill scores during familiarisation would approach overground walking, the difference between overground score and each interval on the treadmill were computed.

Results

Table 1 shows an increase until interval 4, suggesting a relative steady-state performance with ICCs stabilising ≥ 0.86 for the remaining intervals. This is confirmed by the differences from overground walking, which also stabilised from the 4th interval onwards. Interestingly, cadence did not depart considerably from overground values for any of the familiarising intervals, which do not agree with a previous report on healthy subjects, who tended to demonstrate higher cadence at comparable velocities on the treadmill compared to overground [1]. Of the remaining variables, trunk acceleration amplitudes were consistently lower on the treadmill, and so where the regularity variables except ML interstride regularity, which was higher on the treadmill than overground. All these results are in agreement with previous findings for healthy subjects [1]. It should be noticed, however, that performance later in the testing procedure, did not always agree with the familiarisation steady-state scores. Whether this is due to fatigue or other experiences during the testing procedure that followed familiarisation, should not be speculated upon until data collection is concluded.

Table 1: ICC (2.1) between consecutive time intervals during familiarisation, and between familiarisation and 20 per cent test at preferred velocity. Mean values across variables.

Intervals	1-2	2-3	3-4	4-5	5-6	6-7	7-8	8-9	9-10	Familiarisation - 20 per cent test
Cadence	0.75	0.93	0.87	0.98	0.98	0.99	0.85	0.96	0.97	0.57
AP (RMS)	0.65	0.62	0.78	0.87	0.75	0.86	0.93	0.89	0.95	0.82
ML (RMS)	0.83	0.87	0.92	0.88	0.88	0.95	0.95	0.99	0.96	0.81
V (RMS)	0.07	0.80	0.83	0.86	0.84	0.89	0.88	0.88	0.84	0.29
AP Stride Regularity	0.31	0.84	0.82	0.80	0.82	0.72	0.93	0.95	0.84	0.70
ML Stride Regularity	0.47	0.71	0.72	0.76	0.74	0.57	0.77	0.84	0.82	0.60
V Stride Regularity	0.37	0.80	0.84	0.88	0.75	0.62	0.83	0.93	0.79	0.76
Mean across variables*	0.54	0.82	0.84	0.88	0.86	0.87	0.89	0.94	0.91	0.68

* Fisher transformation and re-transformation. RMS: root mean square. AP: Anteroposterior. ML: Mediolateral. V: Vertical

Discussion

The early results from this ongoing study, suggest that ambulating patients post-stroke need a fairly short familiarisation time in order to demonstrate steady-state walking performance on treadmill with BWS.

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A COMPARISON OF FOUR METHODS OF GROUND CONTACT DETECTION FOR NORMAL AND SIMULATED PATHOLOGIC GAIT

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INTRODUCTION

Accurate spatiotemporal gait parameters require accurate foot-ground contact identification. Automated methods can eliminate intra- and inter-rater variability and expedite postprocessing, but these methods have limitations. Force plate data are the gold standard, but are only valid if the foot cleanly strikes the plate.[1] Kinematic methods can provide data for all steps[1], but accuracy and reliability is reduced for variable gait patterns. Footswitches can potentially provide accurate, reliable data for all steps of any gait pattern[2], but the validity of specific systems should be established for specific gait patterns. Here we present a comparison of these three methods and a fourth method based on calculated distance between shoe and ground. The authors evaluated these methods on normal and simulated pathologic gait patterns as an initial validation prior to human subjects testing.

CLINICAL SIGNIFICANCE

Foot-ground contacts must be identified for clinical gait analysis, but all methods do not work equally well for all gait patterns. This study is a comparison of three methods of ground contact detection on four different gait patterns to the “gold standard” of force plate data.

METHODS

Both authors walked normally (Norm) across three force plates while wearing an instrumented shoe on their right foot. One of the authors, a board certified specialist in Neurologic physical therapy with 14 years of clinical experience, additionally simulated normal, shuffling (parkinsonian, PK), ataxic (peripheral neuropathic, PN), and hemiplegic (Hemi) gait patterns. At least 30 right foot initial contacts (ICs) and foot-offs (FOs) were examined for each gait pattern.

Marker data were captured at 120 Hz using a 6-camera Vicon 460 motion capture system (Vicon, Lake Forest, CA, USA). Two AMTI OR6-7 (Advanced Mechanical Technologies, Inc., Watertown, MA, USA) and one Bertec 4060-NC (Bertec Corp., Columbus, OH , USA) in-ground force plates captured ground reaction force data at 1,560Hz. All data were low-pass filtered using recursive Butterworth filters (6Hz for marker and 120Hz for force plate).

Force Plate (FP20N): The vertical component of the force plate data was used to determine initial contact and foot-off when a 20N threshold was crossed.

Kinematic Method (KM): The Vaquita Event Correlator PlugIns (Vaquita, Southampton, UK) were used to define ground contacts for steps based kinematic data from exemplar force plate-defined events.

FootSwitch (FS): The footswitch consisted of six discrete membrane switches (MA-153, Motion Lab Systems, Inc., Baton Rouge, LA, USA) embedded in the shoe sole that communicates with the motion capture system via an active marker.

Digitized Shoe-floor distance (DS): The shoe, eight integral markers, and floor were digitized using a MicroScribe-3DX digitizer (Immersion Corp., San Jose, CA, USA). The floor was digitized using a custom rolling wand. Motion capture data of the shoe markers were used to reconstruct the shoe sole and calculate the minimum shoe-floor clearance for every data frame. The foot was defined to be on the floor when this distance dropped below zero. The delay of the latter three methods were calculated relative to FP20N in ms. A 10N threshold was also evaluated (FP10N) to establish sensitivity to force threshold. Mixed models were used to examine the main effects of ground contact detection method and gait type on deviation of the timing of initial contacts and foot-offs from this gold standard. Mixed models using SAS (SAS Institute Inc., Carey, NC, USA) were used for all statistical analysis.

	Gait	Initial Contact (IC)			Foot Off (FO)		
	Type	KM	FS	DS	KM	FS	DS
Missing	Norm1	-	-	-	-	-	-
	Norm2	-	-	6.7%	-	-	6.7%
	PK	-	2.9%	-	-	-	-
	Hemi	-	-	6.7%	-	-	7.9%
	PN	-	2.6%	-	-	5.3%	-
>3SD from mean	Norm1	-	3.3%	3.3%	-	3.3%	-
	Norm2	-	-	3.3%	-	3.3%	6.7%
	PK	-	-	2.9%	-	5.6%	-
	Hemi	3.3%	-	-	2.6%	2.6%	-
	PN	-	-	-	2.6%	2.6%	5.3%

Table 1. Percentage of data missing and >3SD from mean by method and gait pattern.

RESULTS

Some events went undetected or deviated substantially from the mean **(Table 1)**. Delays in general were greater and more variable for IC than for FO (26±133 vs. 6±51ms). The delay varied by gait pattern and detection method (interaction effect p<0.0001, **Table 2**). Only DS & FS were significantly different from FP20N (p=0.0007 & <0.0001 respectively). FS IC delay was particularly high and variable for simulated pathologic gait patterns.

	Gait	Initial Contact (IC)				Foot Off (FO)			
	Type	FP10N	KM	FS	DS	FP10N	KM	FS	DS
Mean (ms)	Norm1	-1	19	45	7	4	-7	9	43
	Norm2	-1	13	24	9	6	-4	-7	38
	PK	-4	-1	117	10	4	-33	-15	57
	Hemi	-135	-40	378	-77	9	14	-59	98
	PN	-3	25	209	18	10	-32	-34	68
SD (ms)	Norm1	1	10	57	12	1	3	16	5
	Norm2	1	11	29	11	3	5	39	3
	PK	20	60	123	22	2	33	38	59
	Hemi	150	154	356	134	23	24	80	74
	PN	11	17	182	12	6	19	104	45

Table 2. Mean & SD of delay from initial contact & foot-off events relative to 20N force plate threshold by method & gait pattern.

DISCUSSION

We acknowledge that simulating pathologic gait patterns is not ideal, but we believe this technique to be adequate for this initial assessment of the accuracy and repeatability of the methods compared here. Footswitches may have the potential to provide accurate and reliable data for any gait pattern[2], but the footswitches (FS) we tested were not capable of providing this data for all gait patterns evaluated due in part to constraints of their design. Also, while the digitized shoe (DS) method had a significant delay as compared to FP20N in this study; it is much more accurate and reliable for more accurate motion capture systems. These results indicate that any unproven method of ground contact detection should be validated for the specific population and instrumentation to be tested.

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ACKNOWLEDGEMENTS

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LOWER EXTREMITY JOINT KINEMATIC VARIABILITY AS PRODUCED BY VIRTUAL REALITY DURING BACKWARD WALKING

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INTRODUCTION & CLINICAL SIGNIFICANCE

A lot of attention has been given recently to the variability of gait due to its association with gait related disabilities[1]. However, no research has explored how gait variability is affected during more challenging locomotor tasks that has been used for rehabilitation purposes, such as backward walking (BW). During BW there are additional constraints such as the absence of visual feedback and the decreased range of motion Here we investigated how changes in visual feedback, using a virtual reality (VR) environment, affects gait variability during BW.

METHODS

Six healthy young adults underwent three backward and one forward condition on the tread-mill at their self selected pace (SSP) while kinematics was recorded with an 8-camera real-time Motion Analysis system. Subjects walked in a VR environment (Fig. 1) for 8 min at each of the following randomly presented conditions: (1) BW with no optic flow ($BACK_{nVR}$), (2) BW with optic flow perceptually equivalent to their SSP ($BACK_{OFb}$), and (3) BW with optic flow perceptually equivalent to the SSP, but in the opposite direction ($BACK_{OFf}$), and (4) forward walking (FW) with optic flow perceptually equivalent to their SSP ($FORW_{OFf}$). Gait variability was assessed from the lower-extremity joint angles in terms of the sagittal ankle, knee and hip joint's range of motion (ROM) from 350 gait cycles and the stride intervals between two consecutive heel strikes of the right foot. The ROM time series were then evaluated with measures of the amount (CV: coefficient of variation) and the structure (ApE: approximate entropy) of the variability present [1]. Statistical analysis consisted of one-way repeated measures ANOVA.

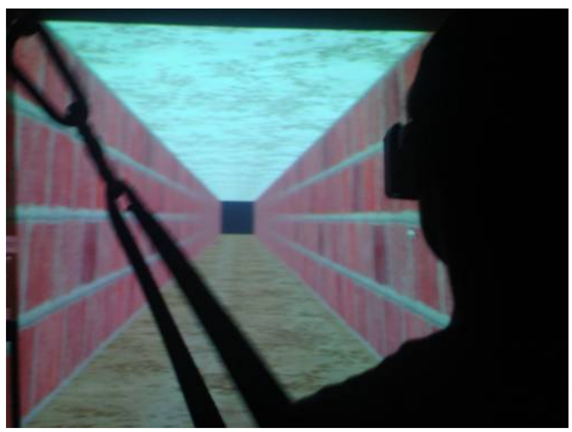

Figure 1. Walking in the VR environment while supported by a vest for safety.

RESULTS

Besides the expected reduction of the SSP (1.75 mph and 2.53 mph for FW and BW, respectively) and the restricted ROM during BW, the CV values were able to differentiate between FW and BW, but found no differences between the BW VR conditions (Table 1). This was also the case for ApE with the exception of the hip ROM where the significant differences were observed between $BACK_{nVR}$ and $BACK_{OFf}$.

Table 1. Group means for coefficient of variation (CV)and Approximate Entropy (ApE) values for the ROM of the ankle, knee and hip joint and stride interval at each condition

		$BACK_{nVR}$	$BACK_{OFb}$	$BACK_{OFf}$	$FORW_{Off}$
Ankle ROM (degrees)	MEAN	24.28*	24.51*	24.43*	33.92*
	CV	10.06*	11.03*	9.89*	5.47*
	ApE	1.173*	1.165*	1.189*	1.489*
Knee ROM (degrees)	MEAN	57.83*	57.07*	60.17*	74.80*
	CV	7.69*	7.60*	7.42*	2.36*
	ApE	1.192*	1.170*	1.159*	1.487*
Hip ROM (degrees)	MEAN	30.03*	29.67*	31.80*	41.16*
	CV	8.29*	8.37*	8.44*	2.47*
	ApE	1.163*‡	1.171*	1.182*‡	1.486*
Stride Interval (sec)	MEAN	1.170	1.162	1.166	1.086
	CV	3.132*	3.354*	3.216*	1.707*
	ApE	1.222	1.184	1.185	1.350

* significant differences between $FORW_{OFf}$ and all backward conditions at 0.05.
† significant differences between $BACK_{OFf}$ and $BACK_{OFb}$ conditions at 0.05.
‡ significant differences between $BACK_{nVR}$ and $BACK_{OFf}$ conditions at 0.05.

SUMMARY/CONCLUSIONS

BW seems to be significantly different than FW with respect to gait variability. However, when optic flow is altered during BW significant differences are not observed in gait variability as it was measured in this study. Since BW is more physiologically demanding [2], it is possible that gait variability has reached a plateau and no further differences can be observed with the measures we utilized. Our results provide fundamental information to understand how optic flow in VR, gait variability, and BW can be utilized for rehabilitation [3].

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FASTING EFFECTS ON PASSIVELY UNSTABLE BALANCE CAPACITY

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Introduction

Poor or compromised balance on passively unstable surfaces can affect performance of many tasks. Occupational settings, however, may require individuals to maintain balance on passively unstable surfaces. Moreover, in a number of work environments the consequences of balance loss or inadequate balance can be especially severe.

While fasting can affect a variety of physical skills and abilities (e.g. load lifting capacity), fasting effects on balance capacity, associated with passively unstable surfaces, are not well understood. Consequently there is a need to investigate these relationships. Consequently, the purpose of the present study was to evaluate the effects of moderate duration fasting on unstable surface balance capacity.

Statement of Clinical Significance

The indication of fasting related effects on passively unstable balance capacity, despite the limited number of subjects suggests that further investigation is warranted. Such investigation should be directed towards populations, such as racehorse jockeys, who perform balance demanding tasks, while often consuming little food.

Methods

The present study sample consisted of eight healthy adult males. Each subject indicated his willingness to participate by signing an IRB approved consent form. Subjects were randomly assigned to either a moderate fasting group or a non-fasting control group. Each subject participated in two testing sessions. The duration between sessions was approximately 6-8 hours for each individual. During the interval between sessions, subjects in the moderate fasting groups were instructed to not consume food or drink, other than water. Control group subjects were directed to consume their customary food and drink.

During each session balance performance was quantified for a wobble board (Fig 1) balance task. Due to the geometry of this instrument, this task involved standing on a passively unstable surface. Balance maintenance on a wobble board required each individual to respond dynamically to the board's movement.

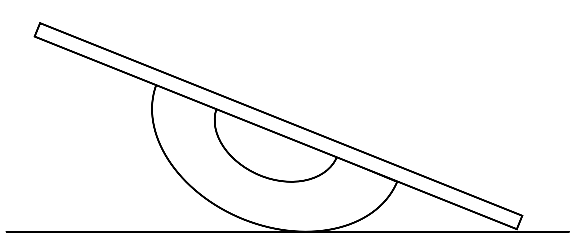

Fig 1. Wobble board

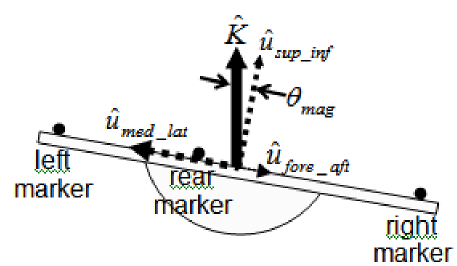

Fig 2. Unit vectors and angles

Each subject was allowed to familiarize himself with the wobble board task, which consisted of standing on the wobble board and keeping the board as still as possible. Following this familiarization each subject was asked to complete three 120 second duration wobble board tasks. Resting time was given to participants between trials. For each trial, coordinate trajectories of wobble board surface markers (left, right and rear edges of wobble board; Fig 2) were measured with an optoelectronic motion tracking system.

Wobble board marker coordinate trajectories were used to compute, at each sampling interval, tilt magnitude angle, θ_{mag}, which represented overall wobble board rotational displacement relative to a horizontal plane (Fig 2). To compute θ_{mag} a unit vector, $\mathbf{u}_{sup\text{-}inf}$, representing a nominal superior/inferior axis (i.e. perpendicular to the board surface) was determined (Fig 2). Recognizing that the scalar product of $\mathbf{u}_{sup\text{-}inf}$ and $\mathbf{K}$ (the global vertical) was equal to the product of their magnitudes and the cosine of the tilt magnitude angle,

$$\mathbf{u}_{sup\text{-}inf} \bullet \mathbf{K} = |\mathbf{u}_{sup\text{-}inf}||\mathbf{K}|\cos(\theta_{mag})$$

and noting that the magnitudes of these vector are unity, θ_{mag} was calculated as

$$\theta_{mag} = \cos^{-1}(\mathbf{u}_{sup\text{-}inf} \bullet \mathbf{K})$$

Results

Changes, from session 1 to session 2, in mean tilt magnitude appeared to be greater for fasting subjects then for control subjects ($p = 0.072$; Fig 3). Similarly, changes in standard deviation of tilt magnitude appeared to be greater for fasting subjects ($p = 0.075$; Fig 3). Neither of these group differences met the traditional a level of 0.05. However, for the small number of subjects in this preliminary effort, the differences were notable.

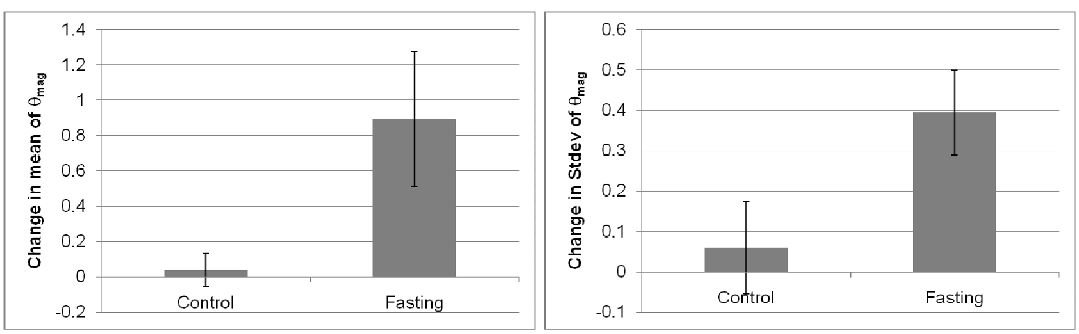

Fig 3. Mean and standard deviation changes in tilt magnitude angle.

Discussion:

While the present study involved pilot data collection with healthy individuals, an occupational population of particular interest is racehorse jockeys. The balance requirements on this population are particularly demanding. There is also a high prevalence within this population (particularly males) of individuals consuming little or no food throughout a racing day in order to maintain body weight below specific levels. Despite the lack of substantive food intake, many jockeys will ride in multiple races throughout a typical race day.

STIFF KNEE PATTERN ALTERS VERTICAL CENTER OF MASS AND LOWER LIMB MUSCLE WORK FOR END STAGE KNEE OSTEOARTHRITIS

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INTRODUCTION

Limited stance phase knee flexion excursion, or a stiff knee pattern, is a feature of the pathomechanics of knee osteoarthritis (KOA) gait. Increasing stance phase knee flexion excursion has been targeted for gait retraining interventions for KOA patients in an effort to normalize gait patterns and to restore efficient gait. However, recent findings questioned the proposed role of stance knee flexion to improve efficiency of walking by decreasing vCOM excursion. Stance phase knee flexion more likely functions as a shock absorption mechanism during weight acceptance. However, this shock absorption role is interrupted by the stiff knee pattern of KOA, such that the knee held in minimal flexion likely serves as a strut to transmit weight acceptance loading through the knee to the hip joint. While it seems plausible that the stiff knee pattern alters the net joint work outputs at the knee, hip and ankle, and perhaps the vCOM excursion, little evidence exists to support these associations. Therefore, the purpose of this paper is to explore the effect of stiff knee pattern on the joint work and the vCOM excursion for subjects with KOA as compared to controls.

CLINICAL SIGNIFICANCE

The effect of the stiff knee pattern of KOA on the vertical COM excursion may help explain compromised concentric limb work during stance phase.

METHOD

Twenty subjects diagnosed with KOA and scheduled for a total knee replacement (TKR) surgery were recruited from a common orthopedic practice and formed the end-stage knee OA group (KOA, 62.6 ± 1.6 years, height = 1.66 ± 0.09; BMI = 32.6 ± 1.08 kg/m^2). Twenty subjects were recruited from the local community to form an age and gender matched control group (CON, 62.7 ± 0.9 years, height 1.70 ± 0.09; BMI = 26.6 ± 0.73 kg/m^2).

All subjects completed standard gait analysis for free speed level walking using an 8 camera motion analysis system, and 2 force plates. OrthoTrak software was used to determine the stance phase sagittal knee position and the sagittal net joint powers for the ankle, knee, and hip. Stance phase knee flexion excursion was determined as the range between the first peak and the minimum prior to contralateral heel strike (Figure 1).

Whole body COM was determined as the weighted sum of the 13 body segment COM positions. The vertical excursion of the COM, defined as vertical separation between the highest and lowest positions, was determined across stance phase for both groups (Figure 2).

The net joint work in the sagittal plane was computed as the area beneath the waveforms of the net joint power. An intra limb combination of joint work was described for the initial period of double stance (WDS_1), for single stance (WSS), and for the terminal period of double stance (WDS_2). The initial double stance joint work was defined as the sum of the hip and knee joint work ($WDS_1 = W_{H1}+W_{K1}$). The single stance joint work was defined as the sum of the ankle, knee, and hip joint work ($WSS = W_{A1}+W_{K2}+W_{H2}$). For terminal double stance, the joint work was defined as the sum of the ankle, hip and knee joint work ($WDS_2 = W_{A2}+W_{K3}+W_{H3}$). The joint work values for WDS_1, WSS, and WDS_2; the stance phase knee flexion excursion and vCOM excursion were assessed for between-group differences using an independent t-test, $P < 0.0125$. Pearson product moments were used to determine the correlations between the stance phase knee flexion excursion and the vCOM excursion and the joint work; as well as, between vCOM and the joint work.

RESULTS

End-stage KOA subjects ambulated with significantly less stance phase knee flexion excursion than age and gender matched controls ($P < 0.000$, Figure 1). When compared to their healthy counterparts, KOA subjects walked with significantly less vertical excursion of the COM during stance phase ($P = 0.001$, Figure 2). No significant between-group difference was seen for joint work during the initial double support phase and for single stance. However, KOA walked with significantly less joint work during the terminal double support phase ($P < 0.000$, Table 1).

Stance phase knee flexion excursion was found to be a strong predictor of vertical COM excursion ($r^2 = 0.45$, $P < 0.000$), and a moderate predictor of joint work during the terminal double stance phase ($r^2 = 0.21$, $P = 0.003$), but was not found to be related to joint work during initial double stance phase and single stance. Joint work during the terminal double support phase was found to be moderate predictor of the vertical COM excursion ($r^2 = 0.14$, $P = 0.019$, Table 2).

DISCUSSION

Subjects with end-stage KOA ambulate with significantly less stance phase knee flexion and vertical COM excursions and these variables show a strong positive correlation (r = .673). Additionally, stance phase knee flexion showed a strong positive correlation to the concentric muscle work of the limb during terminal double support phase (r = .459), which was found to be significantly less for KOA vs. CON. The role of stance phase knee flexion appears to be to increase peak COM position, thereby enhancing the mechanical energy exchange. Stance phase knee flexion excursion also influences the concentric muscle work of terminal double stance, possibly by serving to deliver kinetic energy in much the same way as a recoiling spring. However, subjects with KOA ambulate in response to a pain stimulus at the knee and the functional adaptation of the lower limb during walking should not be altered by increasing knee flexion excursion until the pain stimulus has been relieved

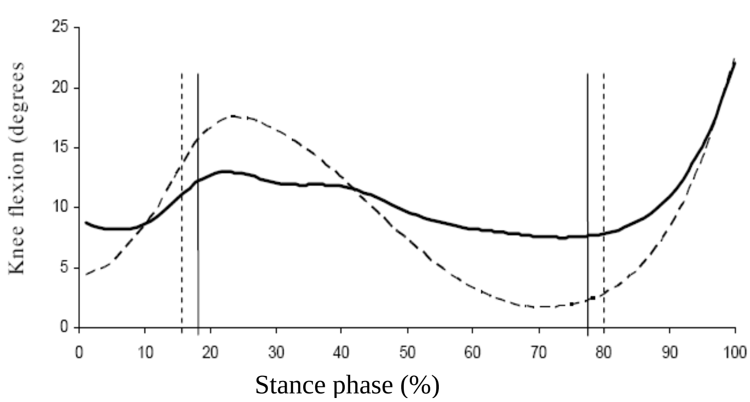

Figure 1. Ensemble average waveforms for sagittal knee position for KOA (solid) and CON (dashed) across stance phase where vertical bars represent contralateral limb toe-off and heel strike for KOA (solid), and CON (dashed).

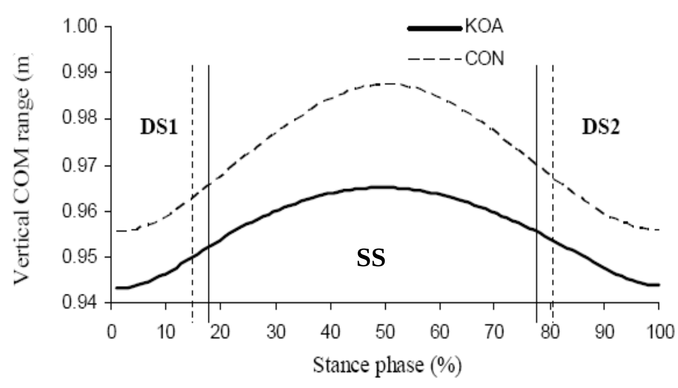

Figure 2. Ensemble average waveforms for vertical COM excursion, defined as the vertical separation between P1 and P2, for KOA (solid) and CON (dashed) across the stance phase. Initial double stance period is indicated by DS1 while the terminal period of double stance is indicated by DS2.

Normative Data of COM Motion in Children, Adolescents, and Young Adults: Balance, Energy Transfer, and Age Differences

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Introduction: Control of the center of mass (COM) motion during walking reflects both whole-body balance and efficiency of energy expenditure. Parameters of COM excursion have been identified as sensitive measurements to distinguish elderly individuals with imbalance from healthy elderly adults [1] and to assess the balance during gait of children with CP [2]. COM movement has also been investigated from the perspective of energy transfer in children with CP [3]. However, there is a need to investigate COM motion from both balance and energy transfer perspectives, and normative data of these COM parameters in children of varying ages are needed. The purpose of this study was to establish the normative values of COM parameters and to investigate whether there are age differences. It was hypothesized that control of COM motion in both aspects of gait balance and energy efficiency would be similar in children, adolescents, and young adults, given that mature gait pattern being developed around age 7 [4].

Statement of Clinical Significance: By establishing the normative data and age differences of COM motion in terms of balance and energy transfer, comparison can be made between children with CP and children with normal development, or before and after intervention. The addition of COM parameters would offer quantitative assessment of balance control and risk of fall in children with CP. It would also provide insights on whole body coordination that can be supplemented to typical gait analyses and ultimately lead to improved treatment planning and better outcome evaluation.

Methods: Data collection from 35 normal subjects between the age of 6 and 26 years (20M & 15 F) were included in this study. Whole body motion analysis was performed when subjects walked at self-selected speed. Full body COM position was calculated as the weighed sum of all body segments (head and neck, trunk and pelvis, upper arms, forearms, hands, thighs, shanks, and feet). Regression equations developed by Jensen [5] were used to define segment mass and position of COM for children (<= 15 years). Dempster's anthropometric data [6] were used for subjects over the age of 15 years. Center of pressure (COP) position was calculated using ground reaction forces and moments measured with two forceplates. Sagittal and frontal COM-COP inclination angles are defined as the angle formed by the intersection of the line connecting the COP and COM with a vertical line through COP [1]. COM potential energy (PE) and kinetic energy (KE) were calculated based on the COM vertical location relative to its mean position and the magnitude of COM velocity [3]. Energy recovery factor and relative phase between PE and KE was also calculated. To compare age differences, the subjects were divided into three age groups (children, adolescents, and young adults), and one way ANOVA was ran to detect significant ($p<0.05$) age differences.

Results: No significant age differences were detected for any of the COM parameters examined (Figs 1 & 2; Table 1). Periods of gait cycle when energy transfer are particularly inefficient are identified in this group of normal subjects (Fig. 2b) and can be used in comparison with data in children with CP.

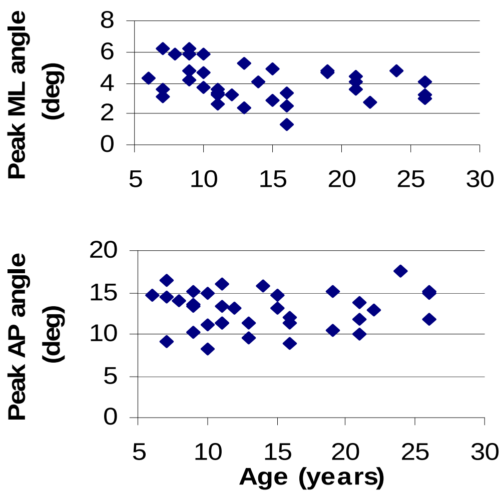

Fig 1. Scatter plots of peak COM-COP inclination angles in the medial-lateral and anterior directions.

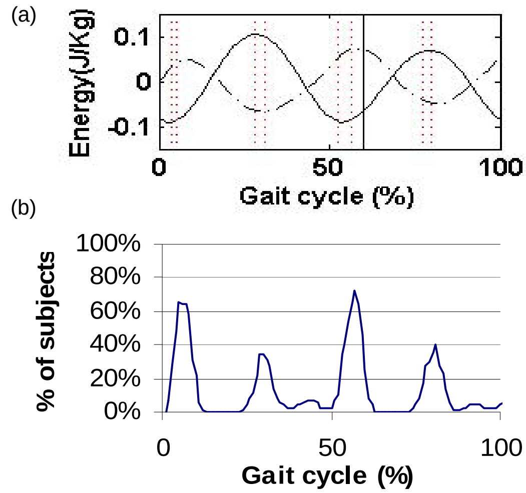

Fig 2. (a) Potential energy (relative to mean COM position, solid line) and kinetic energy (dash-dot) variations about their respective means for a trial. Dotted lines indicate times when PE and KE are in phase; (b) Percent of all subjects showing in phase (inefficient) energy transfer during a gait cycle.

Table 1. Gait velocity, COM balance and energy transfer parameters for each age group. None of the balance and energy transfer parameters showed significant ($p<0.05$) age differences.

	Group1 (6-12y) N = 17	Group2 (13-18y) N = 8	Group3 (19-26y) N = 10
Age (years)	9.2±1.8	14.8±1.3	22.5±2.8
BMI (Kg/m^2)	15.5±2.6	21.0±2.9	24.3±3.3
Gait velocity (m/s)	1.10±0.15	1.02±0.15	1.34±0.16
Peak ML COM-COP inclination angle (deg)	4.38±1.21	3.33±1.33	3.93±0.76
Peak AP COM-COP inclination angle (deg)	12.94±2.39	12.05±2.36	13.32±2.40
Relative phase between potential and kinetic energy (deg)	155.1±19.2	147.6±16.1	158.0±10.1
Energy recovery factor (%)	69.4±8.9	63.6±10.2	65.4±4.9
% gait cycle with in-phase (inefficient) energy transfer	13.9±6.9	17.3±7.8	9.2±5.5

Discussion: To our knowledge, this is the first full-body COM study that examines both balance and energy transfer parameters and establishes the normative data to be compared with patient data. No significant age differences were detected in COM parameters of our subjects, which is consistent with the theory that gait pattern matures at about age seven [4]. Deviations of an individual's data from the normative data might provide an alternative tool to identify deficiencies in balance control or energy inefficiency in gait.

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ASSESSMENT OF DIABETIC NEUROMUSCOLAR DISORDER THROUGH SURFACE EMG

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INTRODUCTION

Diabetic neuropathy (DN) is a neurological disorder caused by consequences of a primary disease which is diabetes mellitus. Among all complications of diabetes, DN can be one of the most debilitating conditions, because of the pain, discomfort, and disability it may cause, and because available treatments are limited and not always successful.
There are three main types of DN: sensory neuropathy or peripheral neuropathy, autonomic neuropathy, and motor neuropathy. Approximately 60–70% of patients with diabetes show signs of neuropathy, but only about 5% experience painful symptoms. According to the categories described above, DN can lead to muscular weakness, loss of feeling or sensation, and loss of autonomic functions. Therefore, methods for early detection of DN are pursued.

CLINICAL SIGNIFICANCE

Early detection of DN may reduce the risk of ulcerations on diabetic foot patients, and so far reduces the number of amputation.

METHODS

Gait and EMG analysis [1-2] were performed with a fullbody markerset [3] on 28 subjects (19 with DN, 9 diabetics (D) without DN, 3 control subjects (C)). BTS motion capture system (6 cameras, 60-120 Hz) and surface EMG (POCKETEMG, 16 channels) synchronized with 2 Bertec force plates (FP4060-10) were used. Surface EMG signals (sEMG) were recorded from peroneus longus (PL), tibialis anterior (TA), gastrocnemius medialis (G), rectus femoris (RF), gluteus medius (GM) and extensor digitorum (ED). sEMG was band pass filtered between 10 and 450 Hz with a zero lag 5th order Butterworth filter and full wave rectified. The rms value was computed with a moving window of 50 ms and the signal was normalized to the mean value in the gait cycle. The gait cycle was determined by using the heel marker trace together with the ground reaction force (GRF) curve. Timing of muscle activation relative to the gait cycle (delays or no activation in "normal" activation regions [2]) were determined, and the amount of EMG activity was quantified as the area under the curve (integral) . The linear envelope of the rectified sEMG was computed by low-pass filtering the signal with a 4^{th} order Butterworth filter and a cut off frequency of 5 Hz. Normalized power was calculated. T-test statistical analysis was performed (see example in Figure 1).

RESULTS

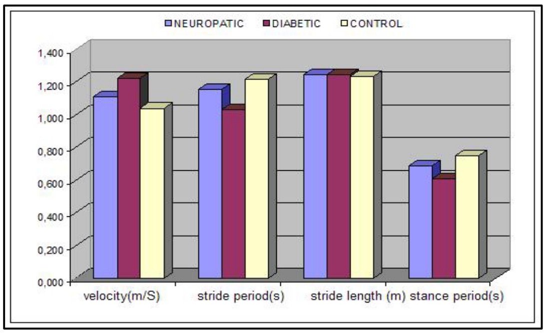

Figure 1: Time and space parameters in DN, D, C subjects

Significant differences in term of integral (Figure 2) and power in DN subjects were found in PL (p=0.00004), TA (p=0.000068), RF (p=0.00005), G (p=0.000002) and ED (p=0.0044) muscles, whereas, in D subjects, they were found in RF (p=0.003), G (p=0.04), and PL (p=0.004) muscles (with respect to C subjects). In term of time and space parameters (Figure 1) significant differences in DN subjects, when compared with C, were found in stance period (p= 0.027). In D subjects and C subjects comparison significant differences were found in stride (p= 0,0001) and stance period (p= 0,000022). Finally when comparing DN and D subjects significant differences were found in gait velocity (p= 0,0001), stride (p=0,003) and stance period (p=0,006).

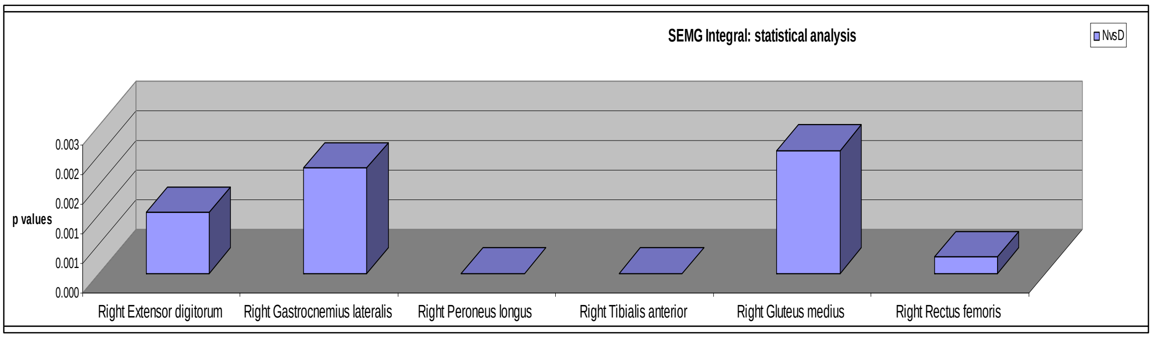

Figure 2: p values of SEMG integral values comparison between DN and D subjects

SUMMARY/CONCLUSIONS

Surface EMG associated with gait analysis showed to be effective detectors of neuromuscular disorders in diabetic neuropathic subjects. Therefore, their employment as methods for early detection of DN is encouraged. Their outcome would also be useful in planning prevention treatments**.** Future developments of the project will consist in extending the study to a larger group of subjects.

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Drop Landing Strategy in Subjects with Recurrent Lateral Ankle Sprains

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INTRODUCTION

Forty-seven percent of ankle sprains occurred in ankles that had been previously sprained. The most common symptoms after ankle sprains were chronic ankle instability, proprioception defect and probable neuromuscular adaptation [1-3]. The purpose of the study was to identify the potential deviations of the landing patterns in individuals with recurrent lateral ankle sprain (RLAS) using detailed biomechanical analysis including analysis of the kinematics and ground reaction force.

CLINICAL SIGNIFICANCE

According the results of biomechanical analysis, we have revealed the adaptation of performing drop landing in the individuals with recurrent ankle sprains. It could be considered as a recommendation of the rehabilitation.

METHODS

Fourteen male adults with RLAS were recruited in this study, and thirteen healthy male adults who were free from any injury of lower extremity were enrolled as the control group. After warm-up, all subjects were asked to perform maximal standing jumps and drop landings from platforms with four different heights (0.37 m, 0.57 m, 0.77m, & 0.97 m). Their kinematics were measured simultaneously by a motion analysis system (VICON 512, Oxford Metrics, UK), and the ground reaction forces (GRF) by two AMTI force platforms. Three successful trials for each condition were performed. Comparisons of the kinematic and kinetic data between the two groups were performed using a t-test with a significance level of 0.05.

RESULTS

During normal drop landing from platforms with different heights, the maximum flexion angles of the hip and knee joints increased with platform height. In the subjects with recurrent ankle sprain, during drop landing from platforms with different heights, the maximum flexion angles of the hip and knee joints, as well as the anterior tilt of the pelvis, hip abduction, and knee external rotation. When compared these two groups, it was found that subjects with recurrent ankle sprain had significantly decreased knee flexion (RLAS: 65.71°±6.43° vs. NORMAL: 70.19°±13.76°) and hip flexion (RLAS: 34.15°±5.42° vs. NORMAL: 42.54°±10.07°), with significantly different times to maximum angles in ankle dorsiflexion and foot pronation. The maximum vertical GRF in the RLAS group was also significantly decreased compared to normal (Fig.1).

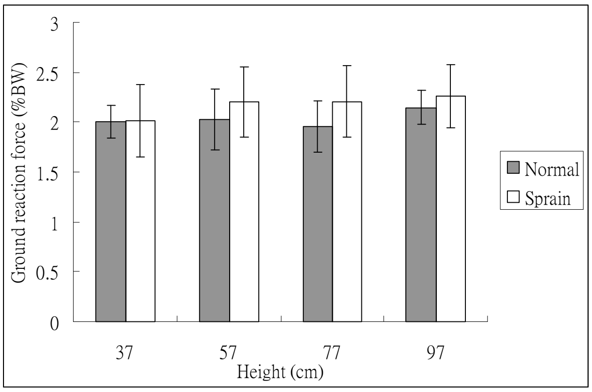

Figure 1: Vertical GRF in both groups for different platform heights.

DISCUSSION

The deviations of drop landing patterns in the individuals with recurrent lateral ankle sprain were identified. Reduced knee and hip flexion adapted in these subjects may be helpful for the reduction of the joint moments when the vertical GRF was increased. The results may be helpful for the rehabilitation of patients with RLAS.

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Stair Negotiation by Healthy Elderly Subjects

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Introduction

As persons age their balance becomes impaired (Lee and Chou, 2007) making the task of descending stairs more difficult. Stair descent has been shown to place large demands on the joints of the lower limbs (Andriacchi et al., 1980; Nadeau et al., 2003). The combination of impaired balance and reduced strength as people age results in a higher incidence of falls causing injury. The kinematics and kinetics of stair negotiation of young and older adults (Stacoff et al., 2005; Mian et al., 2007) has been previously reported. This study aims to add to the pool of knowledge by measuring the motion of the lower limbs of older persons grouped by decade to give a better understanding of the changes that take place as age advances.

Clinical Significance

Understanding the changes in kinematics and kinetics that occur during the difficult tasks of stair climbing and descent may help in the prevention of injury due to falls that are more common amongst the older population.

Methods

Volunteers were recruited from the community for this study and informed consent was properly documented. Subjects were healthy with no known pathologies of their lower limbs. The ninety-two subjects ranged in age from 40 to 86 years. Subjects were divided into groups for the purpose of analysis. Details of the groups are shown in table 1.

Table 1. Gender and age of elderly groups

Group	Male	Female	Total	Age (± SD)
<55	5	10	15	46.8 (5.2)
55-65	5	13	18	60.8 (2.9)
65-75	19	25	44	69.3 (2.6)
>75	12	3	15	78.7 (3.6)
Total	41	51	92	

Motion data was captured using a ten-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA). Patients were asked to ascend and descend four consecutive steps at a self selected speed. The stairs were mounted on two force platforms (AMTI Inc., Watertown, MA) in a manner that allowed three dimensional forces to be recorded from three steps so that reaction forces for each limb were recorded during each trial. A minimum of five trials in each direction were recorded. Data were collected using EVaRT 5 software (Motion Analysis Corp., Santa Rosa, CA) and analyzed using OrthoTrak 6.2.8 (Motion Analysis Corp., Santa Rosa, CA) and MatLab software (The Mathworks Inc., Natick, MA). Statistical analysis was performed using SPSS 14.0 software (SPSS Inc., Chicago, IL).

Results

With the exception of cadence during stair ascent (<55yo - 86.6steps/min; 55-65yo - 77.2; 65-75yo – 76.6; >75yo – 72.2; p = 0.017) and knee power absorbed by eccentric contraction during stair descent (<55yo – 3.8watts/kg; 55-65yo – 3.3; 65-75yo – 3.3; >75yo – 3.0; p =

0.038) (figure 1.), no significant differences were found for the parameters measured using an ANOVA test (α = 0.05) Other parameters showed expected trends when compared to previously reported data for younger and older adults. Notable differences included maximum hip power generated by concentric contraction during stair ascent (<55yo – 3.0watts/kg; 55-65yo – 2.7; 65-75yo – 2.6; >75yo – 2.5; p = 0.080) and maximum vertical ground reaction force (<55yo – 1.4BW; 55-65yo – 1.4BW; 65-75yo – 1.3BW; >75yo – 1.2BW; p = 0.081) during stair descent. Maximum hip abduction and frontal plane range of motion (ROM) during both ascent and descent were similar for all groups except the most elderly which showed reduced motion. All groups exhibited similar pelvic movement in the frontal plane except the youngest group who had a smaller ROM.

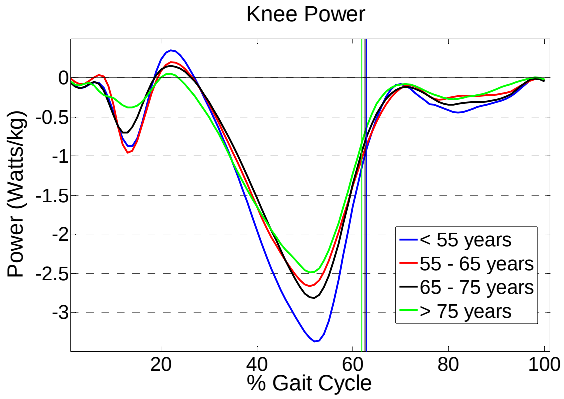

Figure 1. Knee power during stair descent. Negative values indicate movement is and eccentric contraction

Discussion

As persons age their range of motion becomes limited. While negotiating stairs, gross movement of the limb cannot be reduced due to a constrained target for foot placement. Therefore, reductions in flexion angles must be compensated for by larger movement in other planes. As hip flexion reduces with age, subjects need to use greater amounts of pelvis obliquity to move from one step to the next (Lee and Chou, 2007). Analysis of subjects in this study shows an apparent gradual diminution in function for both the knee and hip joint in the sagittal plane with compensations occurring at other joints and in the other planes. The data collected for this study give some indication of the timing and rate at which kinematic and kinetic variables change during the aging process but a much larger dataset is necessary to accurately determine if the changes take place gradually over time or if they occur in a particular order with some changes reliant on other prior alterations in movement pattern. Further evaluation of the cause and effect relationship between parameters such as joint torque and velocity may aid in understanding age related changes in function.

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INCREASING SPINAL RANGE OF MOTION AFTER FUSION SURGERY

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Introduction

One of the more common types of spinal fusion surgeries is performed on adolescents with idiopathic scoliosis (AIS). Critical to this surgery are postoperative changes in trunk range of motion, since the surgery converts a series of vertebral segments capable of inter-segmental motion to a fixed rigid column incapable of motion. Current intuition would suggest that an increase in range of motion (hyper-mobility) between unfused vertebrae above and below the fused section would occur to compensate for the loss of motion in the fused region in these otherwise healthy young patients. This hyper-mobility was thought to be associated with the degenerative changes seen in later years. However, it has been reported that at 1 and 2 years postoperative there is either no change or an actual decrease in range of motion (hypo-mobility) between unfused vertebrae above and below the level of fusion during trunk range of motion tests in AIS patients (Engsberg et al., 2002, Wilk et al., 2006). The purpose of this pilot investigation was to determine if early, postoperative therapy could increase the range of motion between unfused vertebrae following spinal fusion surgery for AIS.

Statement of Clinical Significance

Many patients experience long-term degenerative changes in their unfused vertebral regions. Based upon our results, it is possible that the cause of the degeneration is hypo-mobility. It is also possible that this hypo-mobility can be modified as a consequence of early and/or continued postoperative therapeutic intervention thus improving quality of life.

Methods

Seventeen participants were enrolled and randomly assigned to a control or intervention group. Nine completed the 2-year follow-up testing (6 control and 3 early treatment) (7 females, 2 male; age 15$\pm$1.6 years; height 153$\pm$24.1 cm; mass 56$\pm$9 kg). All patients were treated with an anterior or posterior spinal fusion. The early treatment group was taught an aggressive home therapy program during first postoperative week designed to focus on increasing spinal range of motion. The mobility exercises were done within the painfree range 3 times per week for the first 6 months postoperative. The exercises were performed for: 1) forward and backward flexion, 2) right and left side bending, and 3) right and left rotation. The early treatment group visited a hometown therapist to review the program 2 times per month for the first 6 months. The standard care (control) group was instructed according to typical procedures which were log roll and limited trunk range for 6 months.

Trunk range of motion (ROM) data were collected preoperatively and at 6-12 and 24 months, postoperative using video motion capture. The ROM tests included: 1) right and left lateral flexion, 2) forward flexion, and 3) right and left transverse plane rotation (Engsberg et al., 2002). Three key variables are reported here. The first is the right lateral flexion ROM for segments above the fusion level. The second is the maximum right transverse plane rotation of the shoulders with respect to the pelvis. The third is the maximum sagittal plane forward flexion of the segments below the fusion level. Means and standard deviations were calculated for each group, however tests for significance were not performed because of the small sample size.

Results

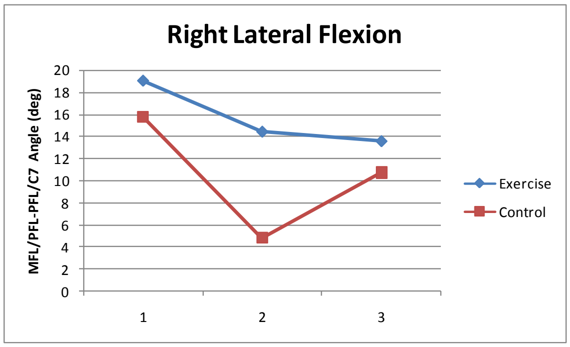

Figure 1. Right lateral flexion range of motion of spinal segments above the fusion level for Early intervention and Control groups over the course of the study. (1-preop, 2-6 to 12 mo postop, 3-2 years postop)

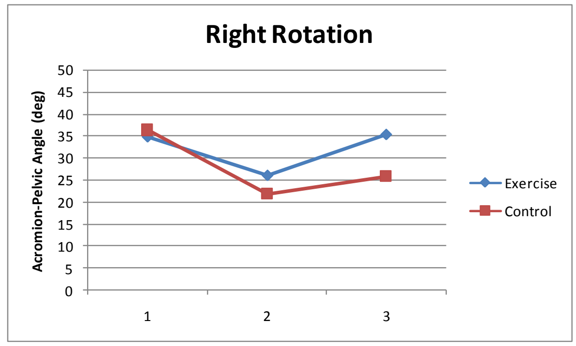

Figure 2. Right transverse plane rotation of shoulder with respect to pelvis for Early intervention and Control groups over the course of the study. (1-preop, 2-6 to 12 mo postop, 3-2 years postop)

Discussion

The major limitation of this investigation was the small sample size, particularly for the Early treatment group (n=3). Nevertheless, it would seem that trends did exist. The Early intervention group had 10° more right lateral trunk flexion above the spinal fusion level compared to the Control at 6-12 months post-op but returned to the baseline difference at 2 years post-op. It would seem that the subjects receiving the early intervention were more willing to move their spine at the 6-12 month post-op testing compared to the control group. The Early intervention and control group had a 5-6° decline between preop and the 2 year post-op test session. The results for right rotation in the transverse plane indicated similar results at the 6-12 month post-op time period as the right lateral flexion results, although not as dramatic (Figure 2). However, the results for the 2 year post-op time period indicated a greater increase, back to baseline values for the Early intervention group compared to that of the control group. The control group declined 6° right rotation from preop to 2 years postop. The results for forward flexion at the unfused region below the fusion level of the two groups are a bit curious (Figure 3) as the groups were not similar at the preoperative time point. The control group clearly had a reduction in range of motion following the surgery. The Early intervention group displayed a dramatic increase at the 6-12 month post-op test session; then returned to a value that was slightly greater than their pre-op values at the 2 year post-op test session. Preliminary results of this investigation support the concept that range of motion at unfused regions can be increased following spinal fusion surgery with an early intervention exercise program.

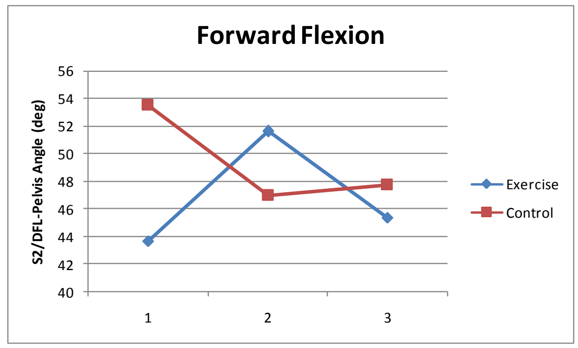

Figure 3. Forward flexion range of motion of spinal segments below the fusion level for Exercise and Control groups over the course of the study. (1-preop, 2-6 to 12 mo postop, 3-2 years postop)

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Wilk et al., 2006. *Spine.* 31:p.309-314.

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Attendees

14th Annual Meeting
Gait & Clinical Movement Analysis Society
GCMAS
Denver COLORADO 2009

Meeting and Committee Schedule

Monday, March 9	Tuesday, March 10	Wednesday. March 11	Thursday. March 12	Thursday, March 13
CMLA (Room 5) 5 - 8:15p	**Executive Board** (Marble Room) 6:30 - 8am	**Executive Board** (Marble Room) 6:30 - 8am	**Executive Board** (Marble Room) 6:30 - 8am	**Educaton Committee** (Marble Room) 6:30 - 8am
	Reimbursement Committee (Sandstone Room) 6:30 - 8am	**Awards Committee** (Marble Room) 3:30 - 4:30p	**CMLA** (Sandstone Room) 6:30 - 8am	**Awards Presented** (Rooms 4-7) 12 - 12:15p
	Educaton Committee (Marble Room) 4:30 - 5:00p	**Standards Committee** (Sandstone Room) 3:30 - 4:30p	**CMLA** (Sandstone Room) 12:15 - 1:15p	
	SMALnet (Limestone) 7:30 - 9:00p	**CMLA** (Sandstone Room) 4:30 - 6:00p		
		Student Gathering (Rooms 4-7) 5 - 6p		
		User Group Meetings (Rooms 5-7) 6 - 8p		

Time scale: 6:30, 6:45, 7:00, 7:15, 7:30, 7:45, 8:00, 8:15, 8:30, 8:45, 9:00, 9:15, 9:30, 9:45, 10:00, 10:15, 10:30, 10:45, 11:00, 11:15, 11:30, 11:45, 12:00, 12:15, 12:30, 12:45, 13:00, 13:15, 13:30, 13:45, 14:00, 14:15, 14:30, 14:45, 15:00, 15:15, 15:30, 15:45, 16:00, 16:15, 16:30, 16:45, 17:00, 17:15, 17:30, 17:45, 18:00, 18:15, 18:30, 18:45, 19:00, 19:15, 19:30, 19:45, 20:00, 20:15, 20:30, 20:45, 21:00, 21:15

14th Annual Meeting
Gait & Clinical Movement Analysis Society
GCMAS
Denver COLORADO 2009

Conference Schedule

Times: 7:00, 7:15, 7:30, 7:45, 8:00, 8:15, 8:30, 8:45, 9:00, 9:15, 9:30, 9:45, 10:00, 10:15, 10:30, 10:45, 11:00, 11:15, 11:30, 11:45, 12:00, 12:15, 12:30, 12:45, 13:00, 13:15, 13:30, 13:45, 14:00, 14:15, 14:30, 14:45, 15:00, 15:15, 15:30, 15:45, 16:00, 16:15, 16:30, 16:45, 17:00, 17:15, 17:30, 17:45, 18:00, 18:15, 18:30, 18:45, 19:00, 19:15, 19:30, 19:45, 20:00, 20:15, 20:30, 20:45, 21:00, 21:15, 21:30, 21:45, 22:00, 22:15, 22:30, 22:45

Monday 3/09/09

- 1-5 Registration (Foyer)
  Exhibitor Setup (Rms 1-3) and A-V setup (Rms 4-7)
- CMLA (Marble Room)

Tuesday 3/10/09

- 7-8 Breakfast (Exhibitor Area)
- 8-10 **Tutorial #1** Ultrasound Imaging of Muscle Deformity in Cerebral Palsy (Room 4-7)
- Break: Snacks in Exhibitor Area
- 10:30-12:30 **Tutorial #2** Musculoskeletal Modeling with OpenSim: (Room 4-7)
- ***Lunch (on your own)***
- 1:30 - 3:30 **Tutorial #3** Computer Adaptive Testing: An Emerging Method of Clinical Outcome Measurement (Room 4-7)
- Break: Snacks in Exhibitor Area
- 4-6 **Research Symposium:** GCMAS Research Directives Planning Initiative (Room 4-7)
- 6:30-7:30 ***Welcome Reception*** (Exhibitor Area)
- 7:30-9:00 **SMALnet** (Limestone)

Wednesday 3/11/09

- 7-8 Breakfast (Exhibitor Area)
- 7-8 **Sunrise #1** SAMSA: Standard Assessment of Motion System Accuracy (Room 4)
- 8-8:15 **Welcome**: J Carollo & Michael Schwartz
- 8:15 - 9:15 **Keynote #1:** What Do Rubber Bands, Penguins and Kangaroos Have in Common? R. Kram, PhD (Rooms 4-7)
- 9:15 - 10:25 **Podium #1** [7 presentations] (Rooms 4-7)
- 10:25 - 11:05 BreakSnacks in Exhibitor Area
- 11:05-12:05 **Podium #2** [6 presentations] (Rooms 4-7)
- 12:05 - 1:10 ***CMLA Accreditation Workshop*** **Box Lunch**
- 1:10 - 2:30 **Podium #3** [8 presentations] (Rooms 4-7)
- 2:30 - 3:00 Break: Exhibit Area & Snacks
- **3 - 4 Poster Odd Attended** (Foyer)
- **4-5 Poster Even Attended** (Foyer)
- 5 -6 ***Student Mixer*** (back of Rms 4-7)
- 3-6 Exhibitors Open:

Thursday 3/12/09

- 7-8 Breakfast (Exhibitor Area)
- 7-8 **Sunrise #2** Surgical Correction of Crouch Gait T. Novacheck (Room 4)
- 8 - 8:15 Buffer
- 8:15 - 9:15 **Keynote #2** Models of Human Gait Using Dynamic Walking Principles A Kuo, PhD (Rooms 4-7)
- 9:15 - 10:25 **Podium #4** [7 presentations] (Rooms 4-7)
- 10:25 - 11:05 Break: Snacks in Exhibitor Area
- 11:05 - 12:15 **Podium #5** [7 presentations] (Rooms 4-7)
- 12:15 - 1:15 ***Business Lunch*** (Room 4)
- 1:15 - 2:45 **Podium #6** [9 presentations] (Rooms 4-7)
- 2:45-7 **Tour Option #1:** Denver Art Museum and WineTasting (same location)
  3-7 Exhibitors Tear Down **(Rooms 1-3)**
- 2:45 - 7:45 **Tour Option #2:** Coors/Red Rocks

Friday 3/13/09

- 7-8 Breakfast (Capitol Foyer)
- 7-8 **Sunrise #3** GPS: A Roadmap to Automatic Gait Data Interpretation R. Baker (Room 4)
- 8-8:15 Buffer
- 8:15 - 9:15 **Keynote #3** Characterizing Dystonia & Spasticity in Neuromuscular Disease T. Sanger (Rooms 4-7)
- 9:15-10:15 **Podium #7a** (6 presentations) (Rooms 1-3)
- 9:15-10:15 **Podium #7b** (6 presentations) (Rooms 4-7)
- 10:15-10:45 Break/Room Change
- 10:45 -12:00 **Podium #8a** [7 presentations] (Rooms 1-3)
- 10:45 -12:00 **Podium #8b** [7 presentations] (Rooms 4-7)
- 12:00 - 12:15 **Awards Presentation** Rm 4-7
- 12:15 - 12:30 Buffer
- 12:30 - 3:00 **Technical Symposium / Box Lunch** Biomechanical Models: As simple as possible, but no simpler (Rooms 1-3)
- Buffer
- 3:15 - 5:45 **Clinical Symposium** Evaluation of Dystonia in the Motion Analysis Laboratory (Rooms 1-3)
- 6:30-7:30 ***Happy Hour / Game Hour*** Foyer & Rooms 1-3
- ***7:30 - 11:30+ Gala Dinner*** Entertainment by Wild 6 (Rooms 4-7)